An investigation of in-shoe plantar pressures and shear stresses with particular reference to diabetic peripheral neuropathy.

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AN INVESTIGATION OF
IN-SHOE PLANTAR PRESSURES AND SHEAR STRESSES
WITH PARTICULAR REFERENCE TO
DIABETIC PERIPHERAL NEUROPATHY

by

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ABSTRACT

Patients with diabetes who also have peripheral neuropathy, frequently develop ulcers beneath the forefoot. It is generally held that the action of repetitively applied stresses during walking is responsible for the development of these ulcers at sites of bony prominences. However, whereas the role of pressure in ulcer formation has been extensively studied, the role of shear has not. Previous attempts at in-shoe plantar shear stress measurement have resulted in unreliable data as a result of inadequate methodologies being employed. In addition, workers have not attempted to relate their findings to the biomechanical function of the foot.

The aim of this study was to investigate shod foot function based on the measurement of in-shoe plantar stresses, with the specific objective of determining the role of pressure and shear in the formation and management of ulcers beneath the insensitive diabetic foot.

Plantar stresses were measured locally beneath the medial four metatarsal heads and heel of the feet of asymptomatic subjects and patients with diabetes who had a history of neuropathic ulceration. In-shoe plantar pressures were recorded using a commercially available insole sensor, while plantar shear stresses were recorded with a discrete bi-axial transducer developed in the Department of Medical Engineering and Physics at King's College Hospital (Dulwich). The simultaneous measurement of mutually perpendicular components of shear stress enabled local resultant maximum peak shear stresses to be calculated.

In contrast to other studies, there was found to be no statistically significant difference in local maximum peak pressures and local maximum peak shear stresses between the two groups ($p>0.05$). However considerable inter-subject, as opposed to inter-patient, variability was apparent in the waveforms of local plantar shear generated; and there was a greater proportion of patient, as opposed to subject shear records, which had bi-directional waveforms, suggesting more movement of the underlying bony structure of the foot or less soft tissue compliance in the patient group, during late stance.

The application of pressure and shear in combination, through actions such as twisting about the forefoot at push-off or anterior movement of the underlying bony structure of the forefoot upon weight-bearing, is considered particularly destructive to subcutaneous tissues through the generation of high internal stresses. Shoewear management to minimise shear in the plantar soft tissues during walking, similar to the presently available methods of minimising local peaks of plantar pressure, may be beneficial in reducing the current numbers of patients with diabetes who develop plantar ulcers.

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CHAPTER 1
INTRODUCTION

The measurement of pressure and shear stresses beneath the plantar surface - the sole of the foot - has been of interest to researchers studying normal and pathological foot function for over a century. During this time numerous ingenious devices for quantifying stresses between the sole of the foot and its supporting surface have been developed and various patient groups studied (reviewed in **Lord, 1981** and **Alexander et al, 1990**). Rheumatoid arthritis, diabetes mellitus, hallux valgus deformity and foot surgery have all been found to affect the plantar pressure distribution in specific ways (**Lord et al, 1986**).

In plantar stress studies, patients with diabetes have been the most extensively studied patient group. The improved diagnosis and better control of diabetes with the use of insulin, controlled diet and other drugs has increased the life expectancy of these patients and raised the potential for the development of foot problems. In the UK there are approximately 750,000 diagnosed diabetic patients, a further 250,000 suspected, and 60,000 new cases diagnosed each year (**British Diabetic Association Report, 1990**). Ninety per cent of the diabetic population is over 50 years old, which is significant in that most foot problems occur after this age (**Levin, 1986**).

Fifteen per cent of the diabetic population are prone to foot ulceration (**Levin and O'Neal, 1988**). Ulcers on the *plantar* surface are typically found in patients with diabetes who also have peripheral neuropathy, a condition characterised by sensory loss, dry skin and small muscle paralysis. Ulcers on the upper, *dorsal*, surface of the foot are typically found in patients who also have a reduced blood supply to their lower limbs, a condition known as *ischaemia* (**Levin, 1986**). Amputation becomes necessary in patients with peripheral neuropathy who develop bone infection through the open site of ulceration, and in patients with ischaemia who develop gangrene because of inadequate blood supply. This thesis will focus attention particularly on patients with diabetes who have peripheral neuropathy.

In a number of plantar pressure studies conducted barefoot, patients with diabetes who had peripheral neuropathy but no plantar ulcers were found to have an altered plantar pressure distribution that resulted in areas of abnormally high pressure (**Stokes at al, 1975; Ctercteko et al, 1981; Boulton et al, 1984; Veves et al, 1991**). Recently a prospective study has confirmed, not unsurprisingly, such areas to be common sites of ulceration (**Veves et al, 1992**). Prolonged periods of moderate stress or short durations of high stress are considered particularly damaging to the insensitive diabetic foot (**Jenkin and Palladino, 1991**).

In contrast to an abundance of plantar pressure data, very little plantar shear stress data is available. In 1963 **Bauman et al** proposed shear stress to be as important as pressure in causing damage to the insensitive foot, but at that time no systems were available for its measurement. The first transducer to measure shear stresses beneath the foot was developed by **Tappin et al (1980)**. Using this device **Pollard (1984)** made barefoot and in-shoe measurements and found abnormally high plantar shear stresses to coincide with sites of previous ulceration. This led him to surmise that high shear stresses were also a determinant factor in the site of plantar ulceration.

The understanding that mechanical factors play a part in ulceration has led to effective management through shoewear therapy. A study in the Diabetic Foot Clinic at King's College Hospital found just 26% of patients using supplied bespoke shoewear fitted with moulded inserts re-ulcerated, compared with 83% who opted to use their own shoewear against advice (**Edmonds et al, 1986**). Moulded inserts were recently shown to be effective in reducing peaks of pressure beneath the forefoot, probably by altering the biomechanics of the foot, as a result of the support provided into the arch (**Lord and Hosein, 1994**).

The present dearth of plantar shear stress data, particularly from beneath the diabetic foot, led to the development of a transducer in the Department of Medical Engineering & Physics at King's College Hospital (Dulwich) (**Williams, 1993**), and has allowed this study of in-shoe plantar shear stresses to be undertaken. Together with the use of a commercially available insole sensor for pressure monitoring, the data gathered directly permits a greater understanding of shod foot function, which is important since most daily activity takes place in some form of shoewear. The data gathered can also be included in finite element (FE) models to analyse the stresses in underlying tissues and bone. This latter approach to the study of ulceration may help to clarify the mechanical mechanisms which influence it.

1.1 AIMS AND OBJECTIVES

The overall aim of this study is to investigate shod foot function in groups of asymptomatic subjects and patients with diabetes who have a history of plantar neuropathic ulceration, based on the measurement of in-shoe plantar pressures and shear stresses. The specific objectives of this study are:

- to develop a methodology for in-shoe plantar shear stress measurement;
- to describe normal shod foot function based on the measurement of in-shoe plantar stresses and the assessment of joint ranges of motion;

and

- to determine the role of pressure and shear in causing plantar ulceration of the insensitive diabetic foot.

1.2 OUTLINE OF THESIS

This thesis is composed of eight chapters. Following the introductory **Chapter 1**, **Chapter 2** reviews the subject areas relevant to the research described in the thesis. This includes a background review with descriptions of: the anatomy of the foot with respect to function; an examination for determining the limits of normal movement at the major joints in the foot; normal gait; the foot-shoe relationship; diabetes and diabetic foot complications; and shoewear management of the "at risk" and ulcerated insensitive diabetic foot. Devices developed for measuring barefoot and in-shoe plantar stresses are discussed in the historical review, along with the clinical findings from barefoot and in-shoe plantar stress studies involving asymptomatic subjects and patients with diabetes.

The instrumentation used to record in-shoe plantar shear stresses is described in **Chapter 3**, ie. the shear stress transducer, its electronic interfacing and the data collection system. A description of the transducer calibration procedure is also given and calibration results are presented.

The clinical trial conducted is described in **Chapter 4**. The criteria for subject and patient recruitment to the trial is presented followed by descriptions of the methods employed to record in-shoe plantar pressures and shear stresses; to assess foot mobility; and to assess sensory loss in the feet of the patients.

The results of the clinical trial are presented in **Chapter 5**. The pressure and shear data gathered is discussed in **Chapters 6** and **7**, respectively. **Chapter 8** concludes the Thesis with a general discussion of the methodology developed and instrumentation used for in-shoe plantar shear stress measurement; the results of the clinical trial; and shod foot function. Finally, the role of shear in causing ulceration of the insensitive diabetic foot is hypothesised.

CHAPTER 2
BACKGROUND AND HISTORICAL REVIEW

2.1 INTRODUCTION

A number of factors will determine the plantar stress distribution during locomotion. The most obvious of these include: articulations of the bony structure of the foot upon weight-bearing; the type of footwear being worn; and the nature and site of any fixed foot deformities or plantar pathologies. In this Chapter, these factors and others are discussed in order to provide an understanding of how barefoot and in-shoe plantar stresses are generated and distributed. The many techniques developed to measure barefoot and in-shoe plantar stresses are also reviewed, and the clinical findings of studies conducted with asymptomatic subjects are discussed. Finally, there is a discussion of diabetes, its associated foot complications, and the findings of plantar stress studies conducted with these patients.

2.2 THE FOOT

2.2.1 definitions of movements

The foot can make six gross movements with respect to the lower leg. During ankle *extension* (dorsiflexion) the foot moves toward the lower leg, while during ankle *flexion* (plantarflexion) it moves away (figure 2.1a). Through the movements of *inversion* and *eversion* the sole of the foot is turned inward and outward, respectively (figure 2.1b). If the right foot is viewed from behind, the heel can be seen to move in an inward direction during inversion and an outward direction during eversion. Finally, the movements of *adduction* and *abduction* respectively move the distal part of the foot toward and away from the centre line of the body (figure 2.1c). These two latter movements of the foot are hardly discernible with respect to the stationary lower leg. Instead, to produce significant adduction and abduction of the foot, the entire leg needs to internally and externally rotate, respectively.

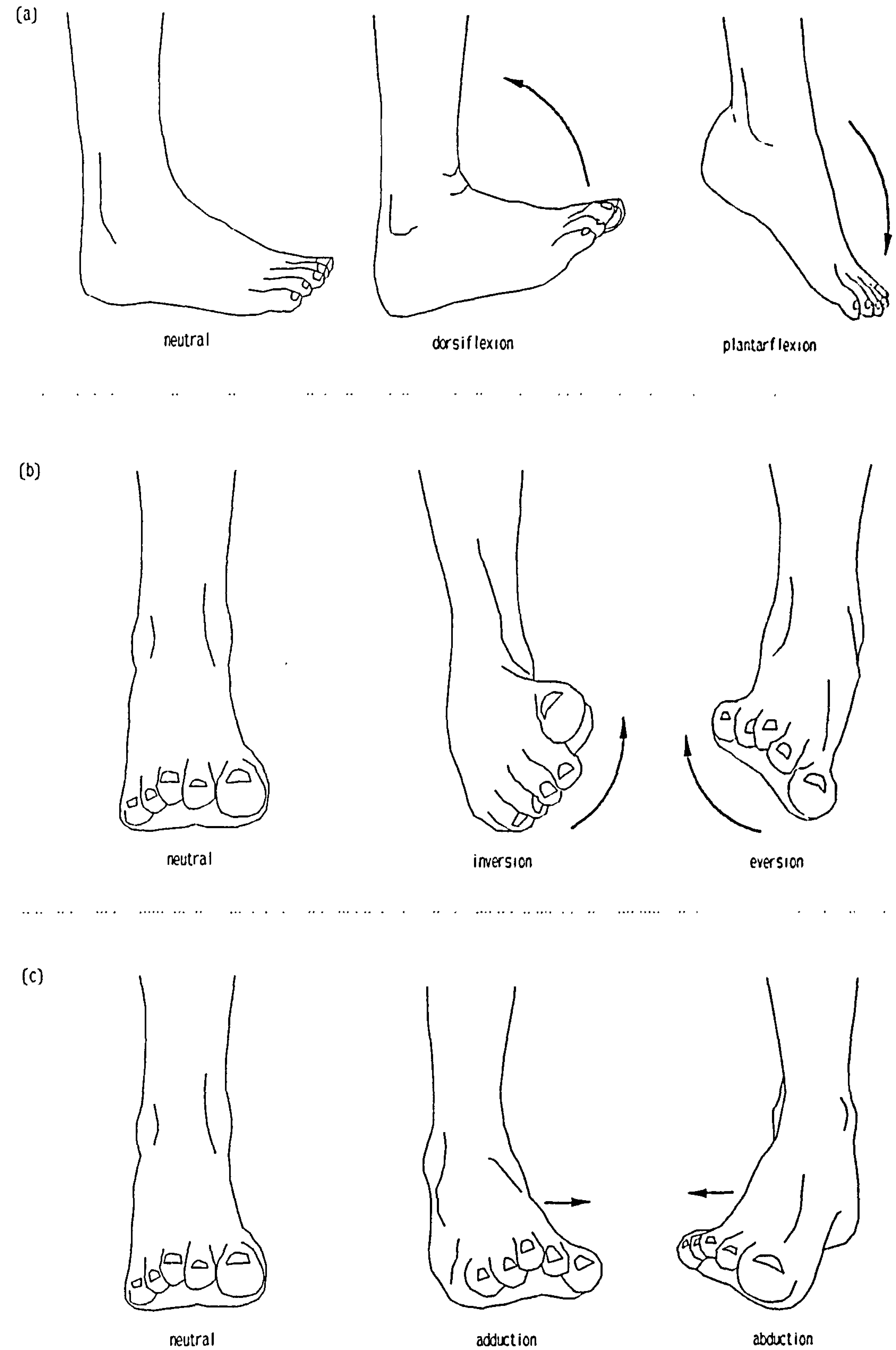

Figure 2.1. Possible movements of the foot with respect to the lower leg.

The movements of the foot on the leg most commonly seen during locomotion typically combine dorsiflexion with eversion and abduction in one direction, and plantarflexion with inversion and adduction in the opposite direction. Some workers refer to the combination of the former three movements as pronation, with supination being the opposite (**Manter, 1941; Wright et al, 1964; Bojsen-Moller, 1979a**). Others define pronation and supination in the same way that eversion and inversion were described above (**Hicks, 1953; Lewis, 1980**). Faced with these and other definitions for the terms supination and pronation (reviewed in **Oatis, 1988**), which essentially describe an inward and outward turning of the sole, **Huson (1987)** put forward a reasoned argument to abandon their use altogether so as to avoid confusion. His argument was based on a comparison of these definitions with the definitions of supination and pronation of the hand, which describe an upward and downward turning of the palm, respectively. To supinate and pronate the hand requires the entire forearm to turn outward and inward, respectively; whereas in the foot, no such movement of the lower leg is necessary to turn the sole inward and outward. In agreement with this argument, the terms supination and pronation will not be used in this thesis. Instead, only the six terms defined at the beginning of this section will be used to describe the gross movements of the foot.

Movements at individual joints are also described by the above terms. For example, the toes move upward and downward during extension and flexion, respectively. The movements of inversion and eversion at individual joints in the forefoot cannot normally be made voluntarily, however, these movements can be made manually by an examiner. For example, if the right foot is viewed from the front, a toe can be inverted and everted by holding it and turning it anticlockwise and clockwise, respectively. In the forefoot, the terms adduction and abduction describe the drawing together and splaying of the toes, respectively. These latter movements occur about a stationary second toe, ie. about the toe adjacent to the big toe.

2.2.2 functional anatomy

The human foot has evolved to serve two main functions: to support the body during standing and to propel the body forward during locomotion. Through evolutionary changes the foot has lost its ability to grasp. The big toe is now no longer divergent and in apposition to the other four toes. Instead, to effect an upright standing posture and bipedal locomotion, the foot has evolved a massive heel, an arched inner border and four lateral toes that are foreshortened to enable the tips to face the ground directly (figure 2.2). In performing its dual functions of pedestal and lever, the foot must at times be rigid and stable, while at others flexible and dynamic. The anatomy of the foot, bone and soft tissue, reflects this duality.

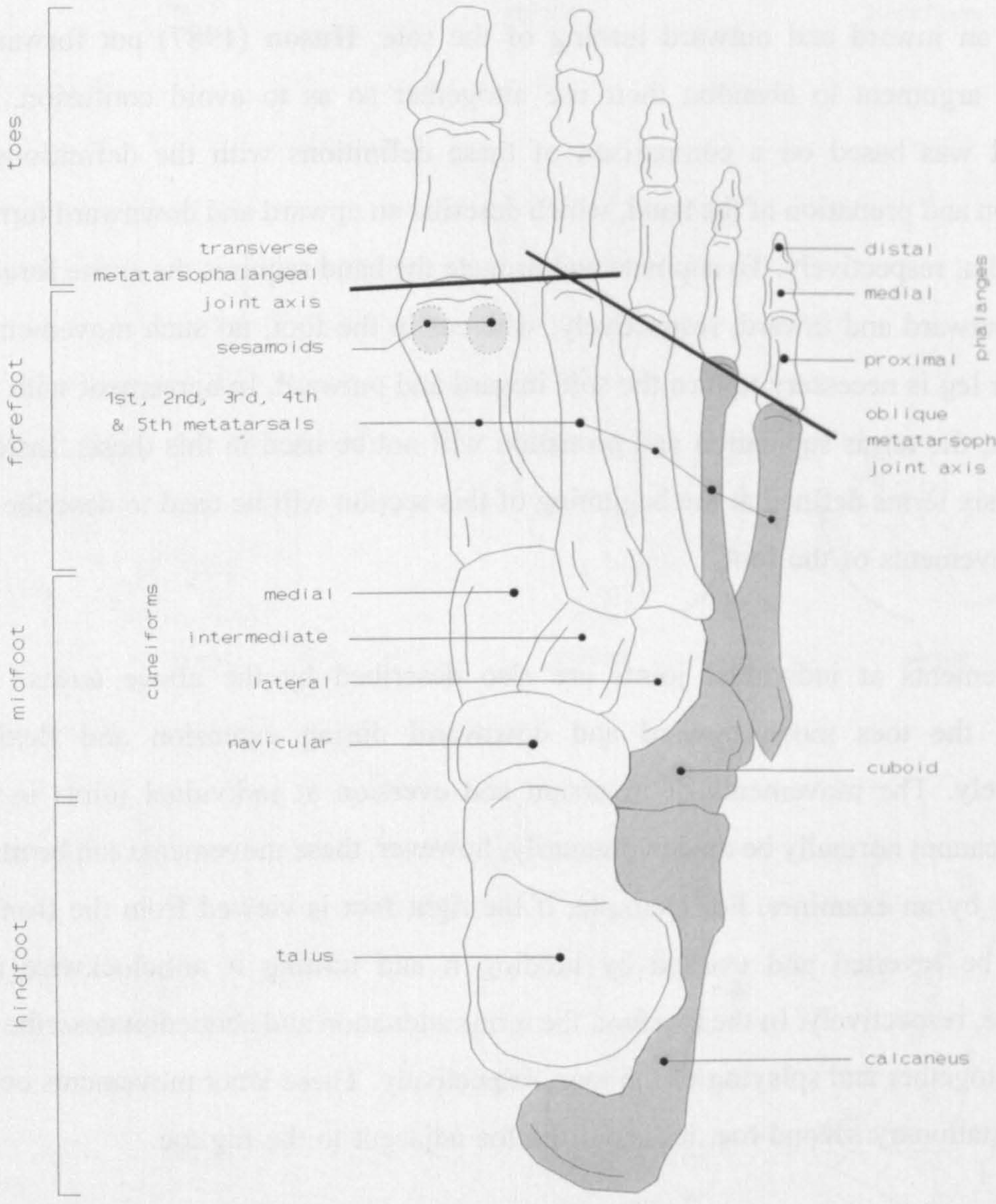

Figure 2.2. Skeleton of the foot in plan view. Shading has been used to indicate the tarsal and metatarsal bones that form the lateral longitudinal arch.

arches of the foot

Three arches are said to span the bones of the foot, two along its length and the third across its breadth (figures 2.3). The *longitudinal* arch is usually divided into two parts: an inner or *medial* part (figure 2.3a) and an outer or *lateral* part (figure 2.3b). Both of these arches are formed by groupings of the tarsal and metatarsal bones. The medial longitudinal arch is formed by the calcaneus, talus, navicular, cuneiforms and the medial three metatarsals; and the lateral longitudinal arch is formed by the calcaneus, cuboid and the lateral two metatarsals (figure 2.2). The *transverse* arch is simply delineated by the relative placement of the metatarsals.

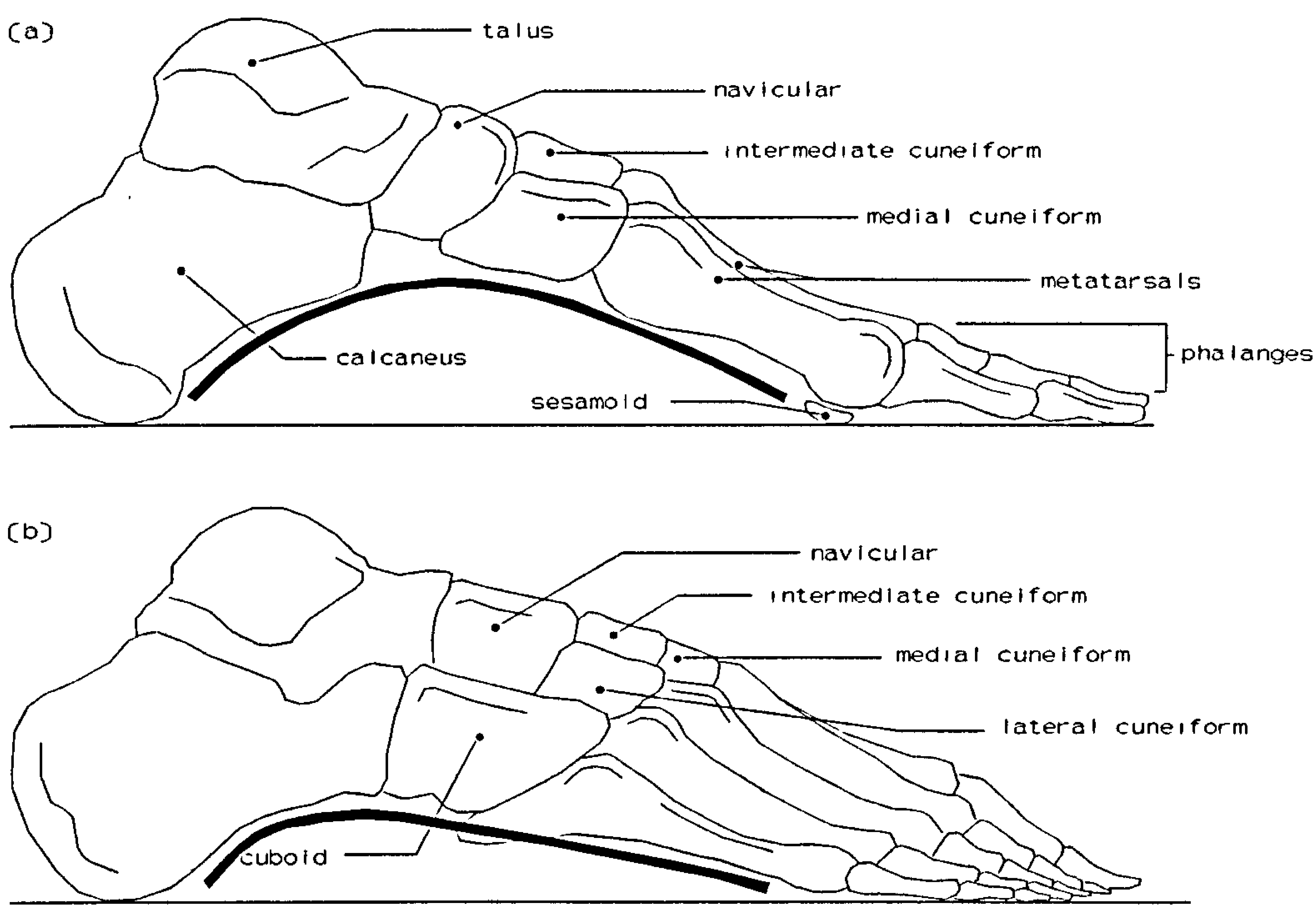

Figure 2.3. Arches formed by the bones of the left foot: (a) medial longitudinal arch (b) lateral longitudinal arch.

When the foot is grounded in static support, the load transmitted downward through the tibia and talus is simultaneously transmitted backward through the calcaneus and forward through the navicular, cuboid and cuneiforms to the metatarsals. The lateral longitudinal arch is normally flattened upon weight-bearing so that the fifth metatarsal lies parallel to the support plane. Flattening of the transverse arch tends to occur at the level of the metatarsal heads (mth) so that all the heads are in contact with the ground. Flattening of the transverse arch at this level has been backed by static plantar pressure studies which have shown the central three metatarsal heads to bear a significant proportion of the total plantar load (**Morton, 1935; Cavanagh et al, 1987a; Hennig and Rosenbaum, 1991; Hughes et al, 1991a**). The medial margin of the foot is not normally in contact with the ground because of the height of the medial longitudinal arch. As a result, this is usually the only arch evident in weight-bearing. The soft tissues beneath the arches that are flattened bear weight, ie. the heel, the lateral side of the foot and the metatarsal heads (the *ball of the foot*). In addition, the distal parts of the toes bear weight.

The medial longitudinal arch is efficiently physically supported by the interaction of the talus, calcaneus and navicular. The talus acts as a keystone, wedged in between the calcaneus and navicular. The lateral longitudinal arch, on the other hand, is more difficult to support because the bones lack a mechanically stable interaction. The studies of **Hicks (1954)** and more recently **Ker (1987)** and **Huang et al (1993)**, have identified the plantar aponeurosis, long and short (calcaneocuboid) plantar ligaments, and the spring (calcaneonavicular) ligament to be the most important soft tissue structures in supporting the longitudinal arches during standing. The plantar aponeurosis attaches to the calcaneus and passes anteriorly, dividing into five slips (bands) which attach to the proximal phalanx of the lesser four toes and the sesamoids beneath the first metatarsal head (figure 2.2). The aponeurosis is more dense beneath the lateral arch in order to compensate for the lack of a stable interaction between the bones (**Napier, 1957**). Muscles that insert into the foot from the leg (the *extrinsic musculature*), and those that originate *and* insert within the foot (the *intrinsic musculature*) have been shown to play minor roles in the support of the arches during standing (**Jones, 1941; Mann and Inman, 1964**).

In standing, the form of the longitudinal arch is maintained through *arch* and *beam* support actions acting together. Arch action results in tension in the plantar aponeurosis and compression forces between adjacent bones; whereas beam action results in bending strain in the bones and tension in the intersegmental ligaments of the plantar surface (**Hicks, 1955**). When weight is borne only on the toes, as in the final stage of foot-ground contact during locomotion, arch action takes over from beam action in the support of the longitudinal arch. As more and more weight is borne by the toes, the distal phalanx slides onto the dorsum of its adjacent metatarsal head and the plantar aponeurosis is pulled forward through a "windlass" action (**Hicks, 1954**). This causes the distance between the end supports of the longitudinal arch to shorten and the arch to raise. Through this process the aponeurosis is placed in tension, the magnitude of which is proportional to the pressure beneath the pads of the toes (**Hicks, 1955**). When the arch is raised to its extreme, arch action assumes the whole load and axial compression in the first, second and third metatarsals reaches its peak (**Stokes et al, 1979**). The windlass action also works in reverse, when the effect of body mass tends to flatten a raised arch. In this case, unwinding the windlass plantarflexes the toes and makes them press more upon the ground (**Hicks, 1954**).

the foot as a lever

The foot functions as a lever to perform its mechanical work. A lever system is made up of three elements: a fulcrum or pivot point; a load which has to be moved; and a force or effort which does the work of moving the load. Based on the relative positions of these elements, three types of lever can be defined.

First order lever. A first order lever is demonstrated by the unsupported foot when it plantarflexes (figure 2.4a). Pivoting takes place at the ankle joint, with the effort produced by the contraction of the *gastrocnemius*, *soleus* and *plantaris* muscles (collectively known as the *triceps surae*), which insert into the calcaneus via the *Achilles tendon.*

Second order lever. The foot functions as a second order lever when it is used to propel the body forward during locomotion (figure 2.4b). This action relies heavily on the contraction of the triceps surae muscles at the back of the leg to enable the weight of the body to be raised.

In comparison to a rigid lever, the foot has a 'break', or flexion axis, where the metatarsals meet the phalanges. At this break the metatarsal heads are not normally in-line, which results in there being more than one axis of dorsiflexion at the level of the metatarsophalangeal joints. These axes are delineated by the relative anterior projections of the metatarsal heads. There are three different forefoot classifications based on the anterior projections of the metatarsals. The most common type is the *index minus*, in which the second metatarsal is projected further forward than the first and the lateral three metatarsals are progressively less anterior (**Viladot, 1991**, see figure 2.2). This is expressed in the following metatarsal formula: 2>1>3>4>5. The other possible types of forefoot are the *index plus*, in which the first metatarsal is projected the furthest forward (1>2>3>4>5); and the *index plus-minus*, in which the first and second metatarsal heads are level (1=2>3>4>5). In a study of one thousand feet, an index minus type forefoot was observed in 56%, an index plus in 16%, and an index plus-minus in 28% (*op cit*).

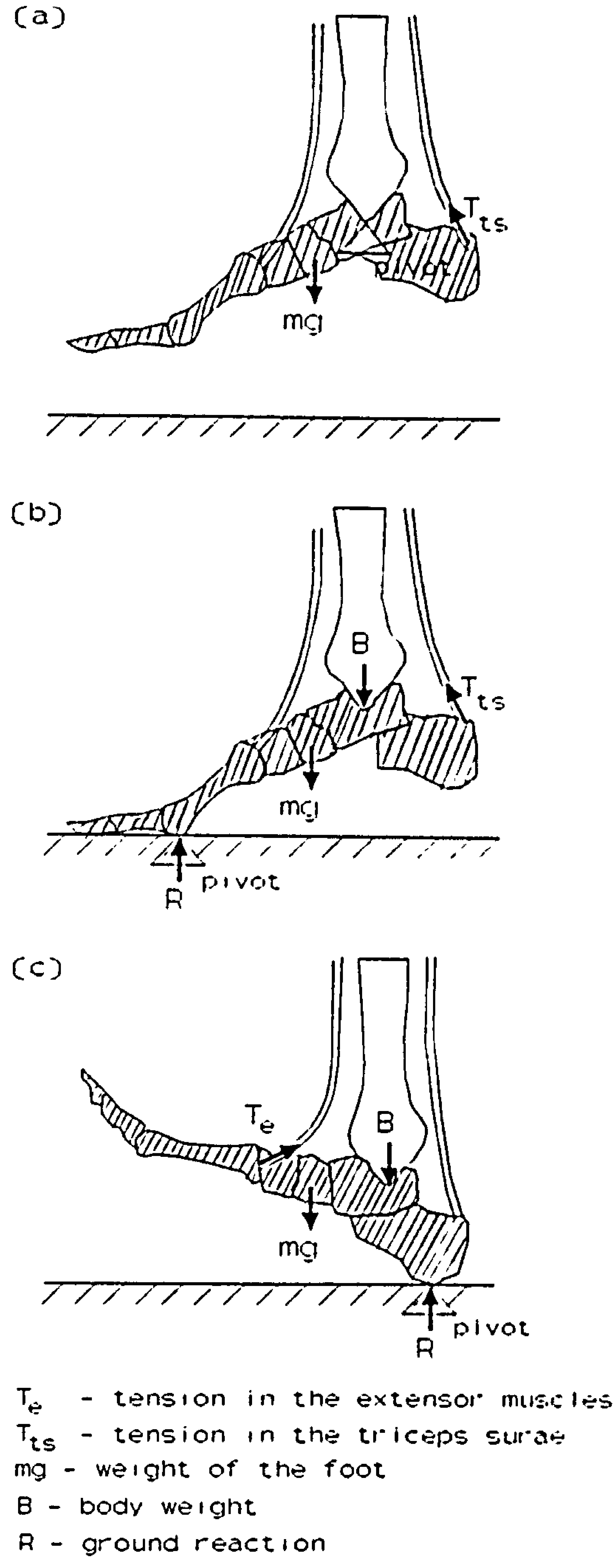

Figure 2.4. Levers of the foot: (a) first (b) second and (c) third order.

The index minus and index plus-minus type forefoot result in two axes of dorsiflexion at the metatarsophalangeal joint level: an *oblique* axis, which passes through the second to fifth metatarsal heads; and a *transverse* axis, which passes through the first and second metatarsals heads. (**Bojsen-Moller and Lamoreux, 1979**, see figure 2.2). In comparison to the oblique axis, leverage about the transverse axis during walking results in increased tension and duration of tension in the triceps surae muscles. Leverage about the transverse axis is further stepped up when the foot pushes off the ground by the strong big toe, which is supported by the tension in flexor hallucis longus muscle. Leverage about the oblique axis, on the other hand, is not as great and further leverage cannot be gained by the lateral toes, which instead simply yield dorsally under load. These two axes can be used for different mechanical demands. The transverse axis is more suited for fast walking, running and sprinting since the necessary leverage can be provided to increase the horizontal speed of the foot as it lifts off the ground; whereas leverage about the oblique axis is more suited for uphill walking (*op cit*).

Third order lever. When the heel makes ground contact during locomotion a third order lever is at work. The anterior extrinsic muscles of *tibialis anterior, extensor digitorum longus, extensor hallucis longus* and *peroneus tertius* are initially contracted to maintain the dorsiflexed position of the foot, but as locomotion progresses they relax slowly to prevent the foot from slapping down onto the ground (figure 2.4c).

the foot as a shock absorber

Shock absorption is the ability of a material or object to reduce the transmitted energy caused by an applied transient force. During locomotion the heel is normally the first part of the foot to make contact with the ground and as a result the soft tissue covering it (the *heel pad*) is also the first in a series of shock absorbers that occur throughout the body. Others include the longitudinal arch of the foot, the knee and the vertebral column. During walking, heel-contact can generate transient forces as high as one-and-a-quarter times bodyweight with frequency components as high as 75 Hz (**Simon et al, 1981**). Shock absorption is therefore necessary to avoid damage to the skeleton, particularly at the joints (**Radin et al, 1972**), and to protect the head from vibration.

The superficial layer of the plantar aponeurosis contains numerous fat-filled septa (compartments) which contribute to the shock-absorbing function of the foot (**Jahss et al, 1992a**). These fibrous septa occur just below the skin and extend from the calcaneus to the metatarsal heads. Impact testing of heel pads has shown the postero-lateral border of the heel pad to provide the greatest shock absorption (**Jorgensen and**

Bojsen-Moller, 1989, Jorgensen et al, 1989). This area has fewer but larger septa and thicker septal walls and is usually the first part of the heel to contact the ground during locomotion **(Bojsen-Moller and Jorgensen, 1991)**. Shock-absorbency of the heel pad is also influenced by its thickness. On average the heel pad is 16.6 mm thick **(Gooding et al, 1985)**. Thicker heel pads may give the advantages of increased time-to-peak-force at heel-contact and reduced peak forces **(Jorgensen and Bojsen-Moller)**, but too thick a heel pad may result in instability during standing and locomotion. The optimum thickness of heel pad results in heel-contact forces being efficiently dispersed in both *space* and *time*: in space by an increase in weight-bearing area beneath the calcaneus as greater load is taken; and in time by relatively slow compression of the heel pad. Fat constituents of the heel pad and the fibrous structures tend to degenerate with ageing **(Jahss et al, 1992a,b)**, which may explain the observed decrease in heel pad shock-absorbency with ageing **(Jorgensen et al)**. However, the shock absorbency of the heel pad may be increased by confinement of the foot in the shoe to reduce spreading of the heel pad under load **(Jorgensen and Bojsen-Moller)**

During rapid compression of the heel pad or of the plantar soft tissues in general, such as during brisk walking or running, part of the impact energy becomes converted into thermal energy **(Thompson, 1983)**. This can be demonstrated as a temperature rise across the plantar surface using thermography **(Brand, 1975)**. As long as there is competent vascularity this thermal energy can be dissipated by the circulation **(Thompson)**. It has been speculated by **Bojsen-Moller and Jorgensen (1991)** that shock absorption is further effected by the squeezing of blood from the vascular bed into the superficial veins, thereby dissipating some of the impact energy for venous return against gravity.

In-vitro drop tests have shown the energy loss in the heel pad to be over 80% **(Jorgensen and Bojsen-Moller, 1989)**. One of the consequences of this in-vitro is that the tendency of the heel pad to recoil at heel-contact is supressed and the heel deforms slowly, staying on the ground to secure grip. On removal of load the original shape is quickly returned, aided by the recoil of the elastic fibres of the septal walls **(Kimani, 1984)** and the low viscosity of the septal fat **(Winter and Reiss, 1991)**.

adaptations of plantar skin to weight-bearing

The skin of the plantar surface has evolved for the demands of weight-bearing and to prevent slippage. Plantar skin is notably thicker than that found elsewhere on the body and is thickest beneath the heel, lateral margin and ball of the foot.

The outer epidermal layer is made up of closely packed cells to form a barrier against chemicals and bacteria. The dermis beneath contains thousands of sweat glands and is transected by a network of blood vessels, lymphatics and nerves. The *sensory* nerves allow sensation of touch, pain and pressure, while the *autonomic* nerves regulate blood pressure and sweating. A high proportion of collagen is also found in the dermis occurring as a network of coiled fibres, while fewer elastic fibres are dispersed throughout. In response to friction and loading the epidermal cells proliferate independently of their natural formation to provide a protective covering of callus (**Naylor, 1955**). Callus is particularly common along the lateral borders of the heel and fifth metatarsal head; and along the medial borders of the hallux (big toe) and first metatarsal head.

The patterns of ridges that are visible on the surface of the skin are in fact created at the level of the dermis in the thin superficial layer. Numerous sweat glands open at the summits of these ridges on the epidermis. When the foot is in contact with a variety of supporting surfaces, the combination of the surface ridges and sweat produced creates a suitable frictional contact to prevent slippage.

behaviour of the plantar soft tissue under load

During standing and in locomotion the soft tissues beneath the foot are subjected to prolonged and repetitive loading, respectively, which can lead to serious tissue damage. The stresses that are applied to the soft tissues during these activities can be resolved into normal and tangential components. The way in which soft tissue behaves under the application of these components of stress has been studied by workers in order to understand the process of soft tissue damage.

Application of pressure. When the foot bears weight the plantar soft tissue thins and bulges outward (**Jahss et al, 1992a**). Within the tissue tensile stresses are set up in the septal walls and the intercellular material (or *ground substance*) slowly squeezes out of the loading site in a time-dependent manner (**Ziegert and Lewis, 1978**). The blood supply is also affected by loading. This is indicated by blanching of the skin, which occurs at low pressures as the result of blood being forced out of the dermis.

Application of shear. Plantar soft tissue subjected to shear will deflect in the direction of shear. The trailing tissue behaves in a similar manner as if it were placed in tension. Initially it exhibits a lax phase during which the collagen fibres straighten and align in the direction of the load, and thereafter the straight collagen fibres impart a high stiffness to the tissue (**Barbenel et al 1978**). Beyond a certain level of shear, however, the soft tissue will not be able to support a shearing load and will tear.

The internal stress distribution of soft tissue subjected to skin shear was investigated by **Zhang and Roberts (1993)** in a theoretical analysis. Skin shear stresses were predicted to alter not only the internal shear stress distribution, but also the internal normal stress distribution; and to have the same effect on the internal stress distribution as pressure. However, this is contrary to **Bennett et al (1979)** who had earlier shown pressure applied to the skin to be roughly twice as effective as skin shear in reducing tissue blood flow.

Application of pressure and shear. During activity the foot is simultaneously subjected to pressure and shear. The simultaneous application of these stresses is thought particularly destructive to soft tissue in that the magnitudes required to produce damage are reduced when they are applied concurrently (**Bennett et al, 1979**). A theoretical analysis of soft tissue, which was initially compressed and then subjected to surface shear, has shown the effects of these stresses to be additive (**Zhang and Roberts, 1993**, figure 2.5). The tissue model showed an internal stress distribution with three distinct zones. Directly underneath the area of compression was a region in which there was no change in the internal stress distribution. However, in the regions ahead of and behind the point of shear application there was an increase and decrease, respectively, in both the internal shear and normal stress distributions.

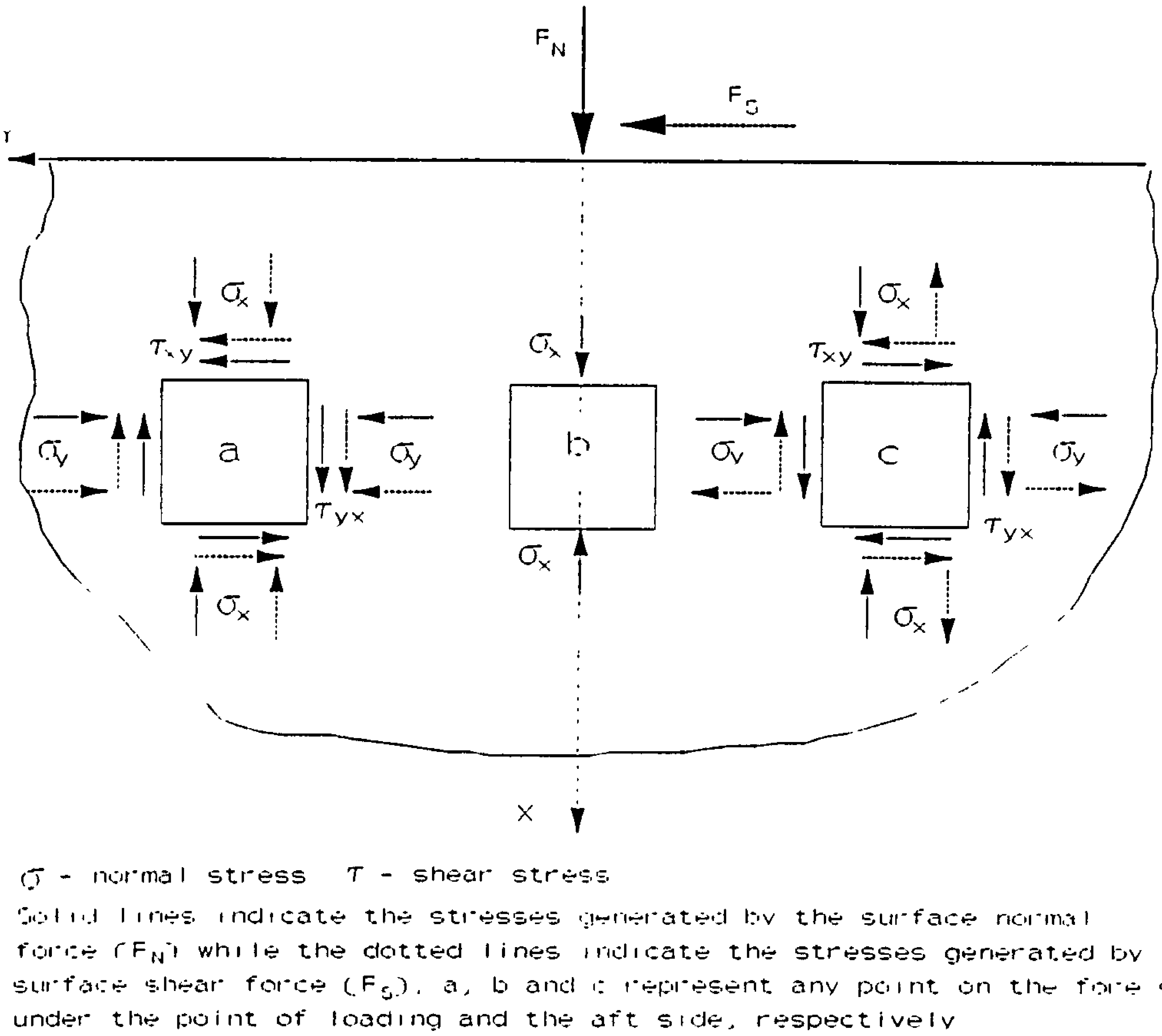

Figure 2.5. Stresses within soft tissue produced by surface normal (F_N) and shear forces (F_s) (from Zhang and Roberts, 1993, reprinted with permission).

mechanisms of plantar soft tissue damage from mechanical insult

The mechanical damage of plantar soft tissue can result from trauma, as in the case of a sharp object penetrating the skin; or instead from a non-penetrating stress that is applied to the skin surface. In this latter case, damage to susceptible tissue can occur from either an apparently moderate stress that is applied over a prolonged period; or from a high stress that is repetitively applied over a relatively short period. These three processes of soft tissue damage are discussed in detail below.

Tissue damage resulting from trauma. The process of tissue damage from a penetrating object is obvious and can occur to almost anyone. However, particular patient groups may be more at risk, for example those with sensory nerve damage that results in anaesthesia of soft tissue.

Tissue damage resulting from prolonged loading. Tissue damage to the feet as a result of prolonged loading usually occurs to those patients who are immobilised through paralysis or disability (physical and/or mental) and are bedridden. The heels of patients with *Spina Bifida* or *Cerebral Palsy* can be affected in this way. Many of these patients can feel pain from prolonged lying, but are unable to make voluntary movements to relieve areas of uncomfortable pressure, which may be indicators of impending tissue damage. Tissue death, or *necrosis*, is known to result from moderate stresses impeding the blood supply and lymphatic drainage for prolonged periods (**Bader, 1990**), but there are also other unknown factors that make patients susceptible to this form of tissue damage (**Meijer et al, 1989**). The ulcers that form in this way are referred to as *decubitus ulcers* or *pressure sores.*

Tissue damage resulting from a repetitively applied load. Damage from repetitively applied stresses most commonly occurs to specific patient groups when other predisposing factors are present. Patients with *Hansen's disease* (*Leprosy*) or *Diabetes Mellitus* who also have sensory nerve damage, particularly in their feet, are commonly affected. Tissue damage through this process is less well understood, but **Bauman et al (1963)** suggest the process begins in susceptible tissue when apparently normal plantar pressures, generated during walking or running, are high enough to crush cells and rupture capillaries. If the foot is rested, the tissues so affected are repaired by scar. However, with continued weight-bearing activity, further damage occurs resulting in an accumulation of necrotic tissue, which eventually ruptures through the skin to form an ulcer.

2.2.3 biomechanical examination of the foot

Normal foot function is partly defined with respect to the ranges of motion that are available at certain joints in the foot. To determine these ranges of motion, a *biomechanical examination* is conducted during which the foot, or parts of it, are manipulated and angular measurements are made at certain joints using a *goniometer* (see figure 2.8). In addition to these measurements, the mobility of the forefoot and the nature of any fixed positions in the feet are also assessed as part of the examination.

The first part of the examination takes place while the patient is lying prone on an examination couch, and starts with the feet and half of the lower legs overhanging the edge of the couch. In this position the legs will naturally take up an externally rotated position with the feet abducted. However, to conduct an examination of the *right foot*, it is first necessary to position the posterior surface of the right heel to lie in the horizontal plane. To do this, the left leg must be flexed at the knee, externally rotated, abducted at the hip, and the left foot placed across the back of the right knee (figure 2.6).

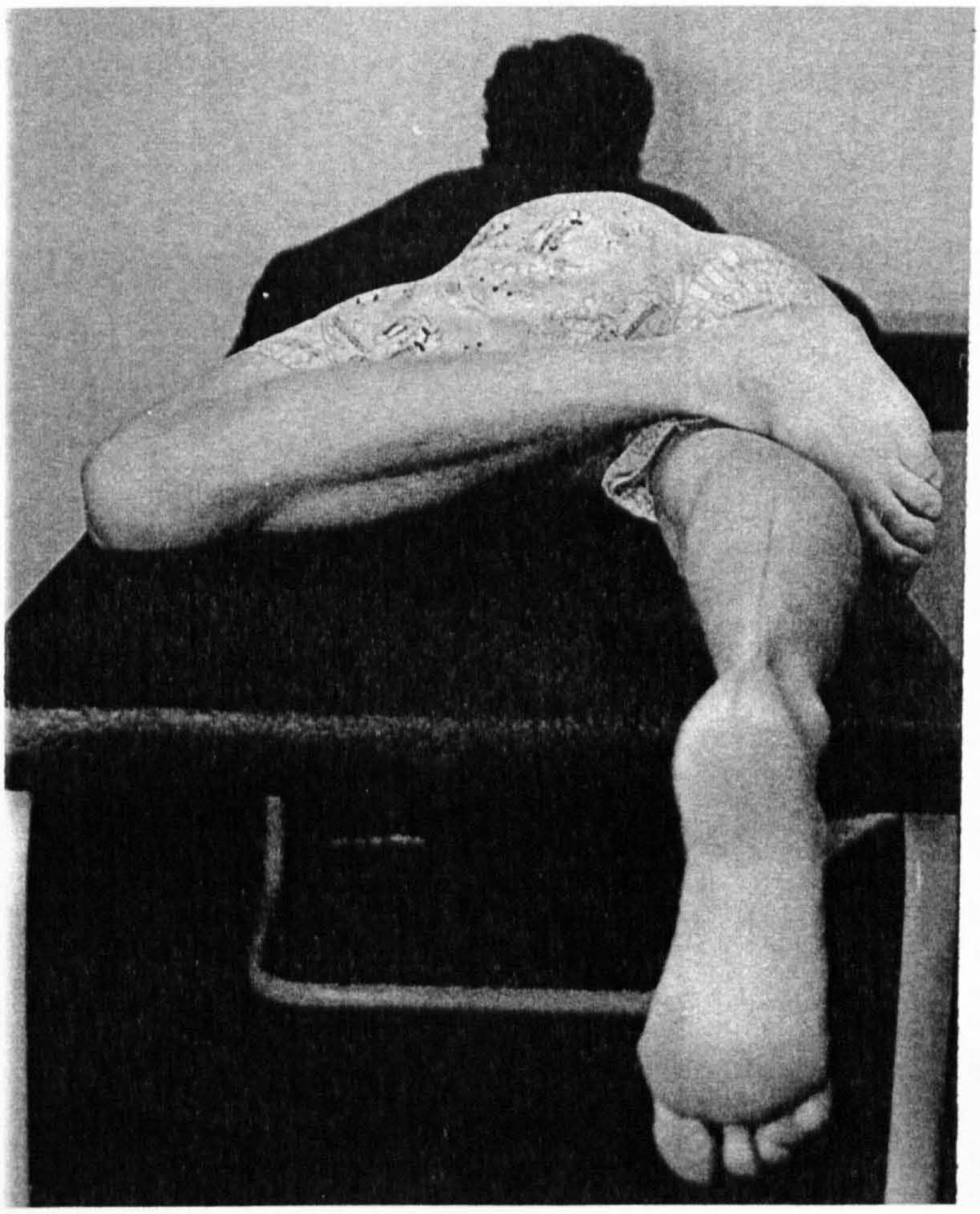

Figure 2.6. Subject in the examination position for the biomechanical assessment.

As an aid to making measurements during the examination four reference lines are drawn on the skin of the foot and leg using a watersoluble marker pen. The first line is drawn along the length of the fibula on the lateral side of the leg, from the lateral malleolus to the fibular head; the second line is drawn along the lateral border of the foot, parallel to the plantar surface (figure 2.7); the third line is drawn along the centre of the posterior surface of the heel; and finally the fourth line is drawn along the *midline* of the leg (figure 2.6). The midline of the leg is found by palpating the gastrocnemius muscle with the middle finger. The finger is pressed into the centre of the back of the leg on the Achilles tendon and moved up towards the back of the knee. The point where the finger meets with resistance, between the medial and lateral heads of the gastrocnemius muscle, is marked and a line drawn from here to another point marked on the back of the leg midway between the two malleoli (figure 2.6). The midline of the lower leg is independent of the Achilles tendon (**Elveru et al, 1988b**), but it should join up with the centre-line of the heel for the two to be in a *straight* line.

Figure 2.7. Method of placing the right foot in the subtalar neutral position.

The ranges of motion that are available at the joints of the hindfoot are measured by manipulation from a starting point termed *subtalar neutral* (or just *neutral*). To place the foot in subtalar neutral the foot must be held with both hands. For the right foot, the head of the talus (found on the anterior surface of the lower leg at the level of the malleoli) is palpated with the left index finger and thumb; and the forefoot is held with the right hand, with the thumb placed over the plantar aspects of the fourth and fifth metatarsal heads (figure 2.7). By inverting and everting the foot with the right hand, the lateral and medial aspects of the talar head alternately become prominent, respectively. When these aspects are felt equally, the forefoot is gently dorsiflexed to meet resistance and into the neutral position. In this position the foot is "locked" in the mortice created by the lower ends of the tibia and fibula; the forefoot is fully everted about the midtarsal joint and "locked" on the hindfoot; and the angle between the lateral border of the foot and the long axis of the leg (indicated by the reference line on the lateral side) should be approximately 90^0 (figure 2.7).

Ankle dorsiflexion is measured from the neutral position. The palm of the hand is placed up against the sole of the foot and the foot is gently pushed into dorsiflexion as far as it can go. With the knee fully extended, 12.6^0 of dorsiflexion is considered normal (**Boone and Azen, 1979**).

Calcaneal inversion and eversion are also measured from the neutral position. The heel is cupped in one hand and fully inverted and everted (figure 2.8). At these extremes, the angle between the centre-line on the heel and the midline of the leg are measured: 36.8^0 of inversion and 20.7^0 of eversion are considered normal (**Boone and Azen, 1979**). As a general rule, there is roughly double the range of inversion as there is eversion in an asymptomatic foot (**Foulston, 1987**).

The ranges of motion measured at the joints of the hindfoot vary widely from study-to-study. This may be due to variations between subjects. For instance, variations in the lengths of lower limb muscles; the interaction of bones; or the angulation of the imaginary axis about which a movement takes place. Ranges of motion may also be influenced by inter-observer reliability (**Elveru et al, 1988a**). For instance, errors may be introduced into measurements when locating and marking reference lines on the foot and leg and/or in aligning the goniometer on these lines.

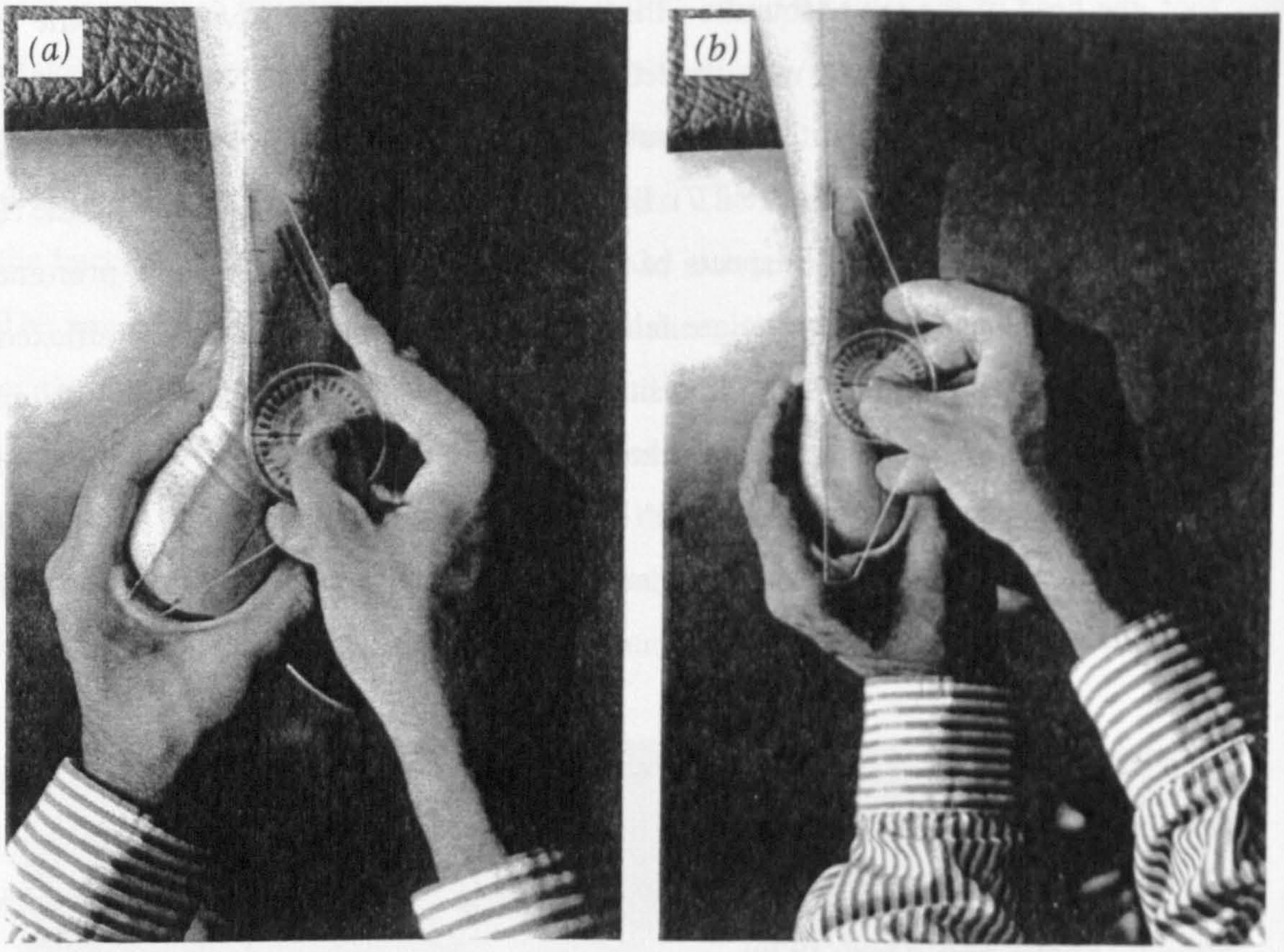

Figure 2.8. Measuring the range of (a) calcaneal inversion and (b) calcaneal eversion of the right foot (nb. the range of calcaneal eversion was abnormally low in this subject.

The metatarsals are assessed for their mobility relative to each other. Only the first and fifth metatarsals are of interest because these are considered to be independent of the central three (**McCourt, 1983**). The firm proximal articulations of the middle three metatarsals only allows them to move in a dorso-plantar direction, and only to a small extent (**Neale, 1981**). The mobility of the first metatarsal is assessed relative to the other four. The forefoot is held in both hands: the left thumb is placed over the plantar aspect of the first metatarsal head and the index finger is placed over the dorsal aspect; with the right hand, the thumb is placed over the plantar aspect of the lesser four metatarsal heads and the index finger is placed over their dorsal aspects. The lesser four metatarsal heads are supported in the same plane with the thumbs placed parallel and the nails level. The first metatarsal is then plantarflexed and dorsiflexed and the difference in the movement of the thumb nails judged at the extremes of movement (figure 2.9). The movement in either direction should be equal - about 5 mm (**Weed**

and Holt[1]). If there is more plantarflexion than dorsiflexion, then the first metatarsal head will lie lower than the others when the foot is in the neutral position, ie. there is a *plantarflexed first metatarsal.* If, however, there is more dorsiflexion than plantarflexion, then the first metatarsal head will be higher than the others when the foot is in neutral, ie. there is a *dorsiflexed first metatarsal.* A similar procedure is carried out to assess the motion of the fifth metatarsal, which is plantarflexed and dorsiflexed against the other four.

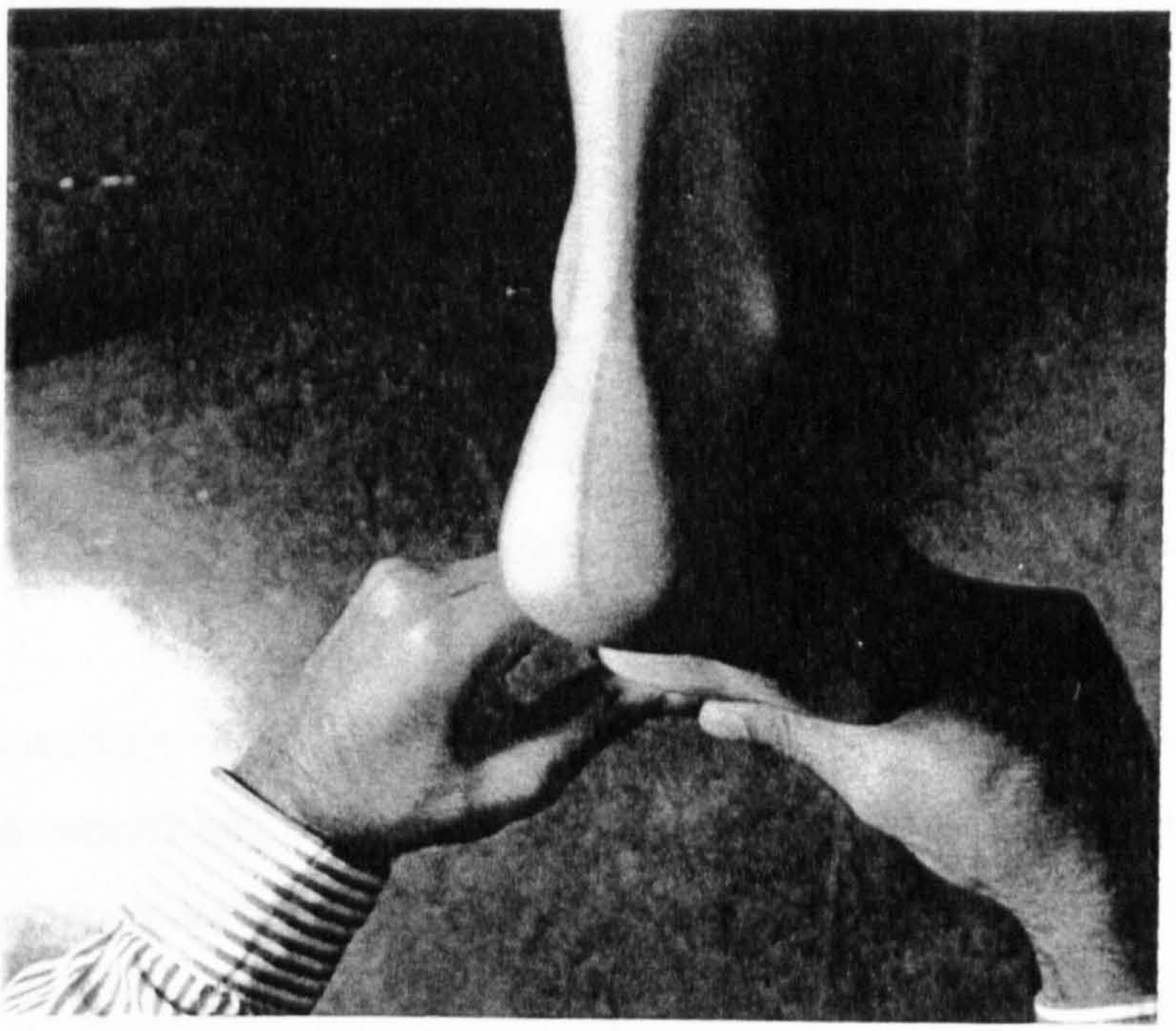

Figure 2.9. Method of assessing the mobility of the first metatarsal of the right foot.

Frontal plane deviations of the forefoot are assessed by looking down on the forefoot from above (figure 2.10). From the neutral position, one arm of the goniometer is aligned perpendicular to the centre-line drawn on the heel and the other arm is placed parallel to the plane of the metatarsal heads. If the first metatarsal head is more dorsal than the fifth, then there is a *forefoot varus*; however, if the fifth metatarsal head is more dorsal than the first then there is a *forefoot valgus.* With a forefoot valgus, the metatarsal heads may all lie in one plane or the first metatarsal

[1] Biomechanics II Syllabus, California College of Podiatric Medicine, 1210 Scott Street, San Francisco, CA 94115.

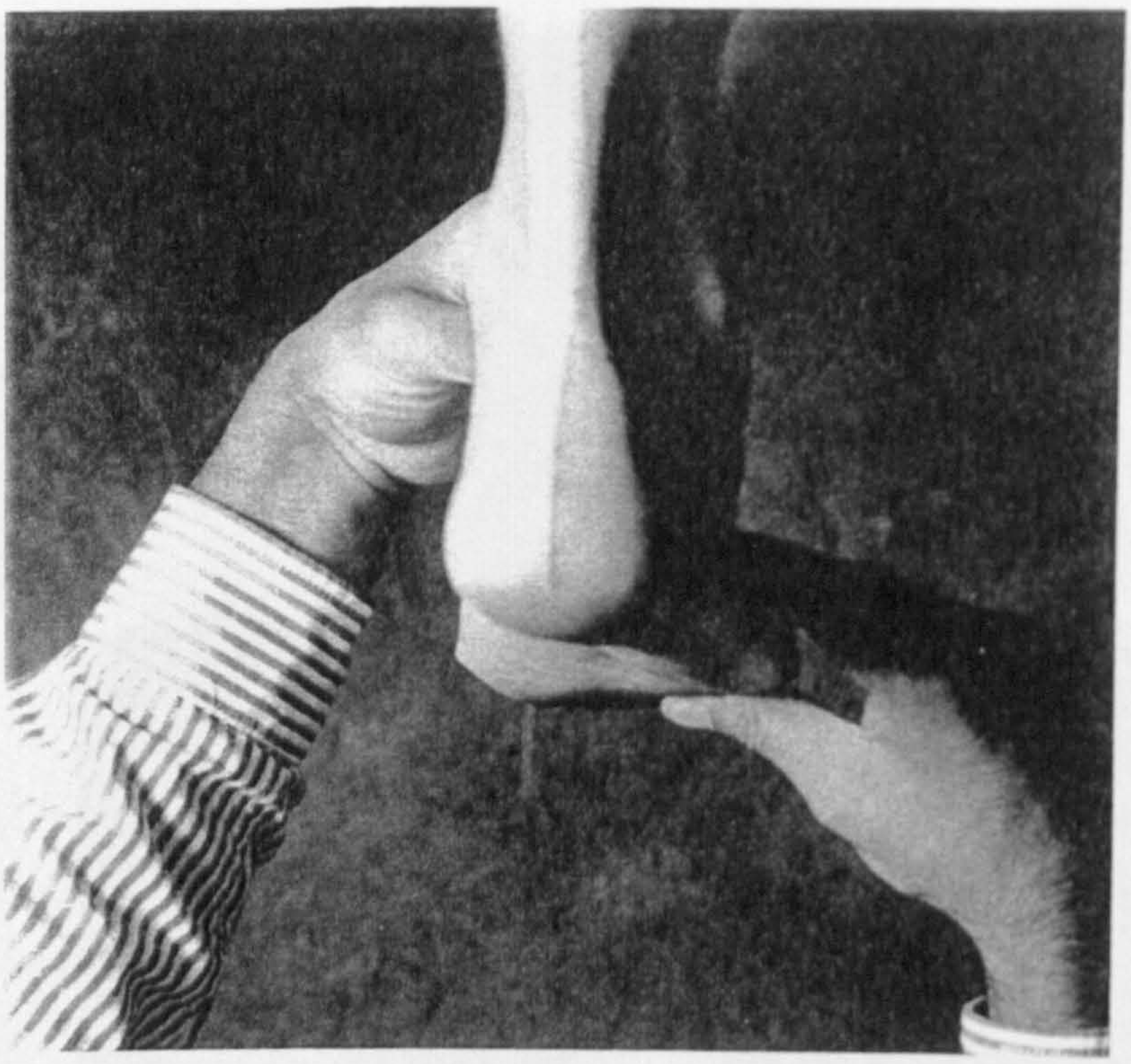

Figure 2.10. Assessing frontal plane deviations of the right forefoot. This subject has a total-type forefoot valgus.

head may be plantarflexed below the level of the others. The former deviation is known as a *total-type forefoot valgus*, while the latter deviation is known as a *plantarflexed first-metatarsal-type forefoot valgus*.

Frontal plane deviations of the rearfoot (calcaneus) and tibia are assessed while the patient is standing in their *angle* and *base of gait*. To achieve this position, the patient starts off by walking up and down on the spot and is then unexpectedly asked to stop. The base of gait is the distance between the heel centres and can be measured from an outline of the feet. The angle of gait is the angle between the midline of the feet. The midline of the foot is an imaginary line drawn on the sole, from the centre of the heel and through the second metatarsal head. In practice, the angle of gait is estimated by placing rulers up against the medial borders of the feet and using the goniometer to measure the angle in between (figure 2.11). Standing in their angle and base of gait, the angle between the centre-line of the heel and the horizontal determines whether there is a *rearfoot varus* or *rearfoot valgus* condition (figure 2.12a). Normally this angle is 90^0. If it is greater than 90^0, as measured on the medial side of the foot, then

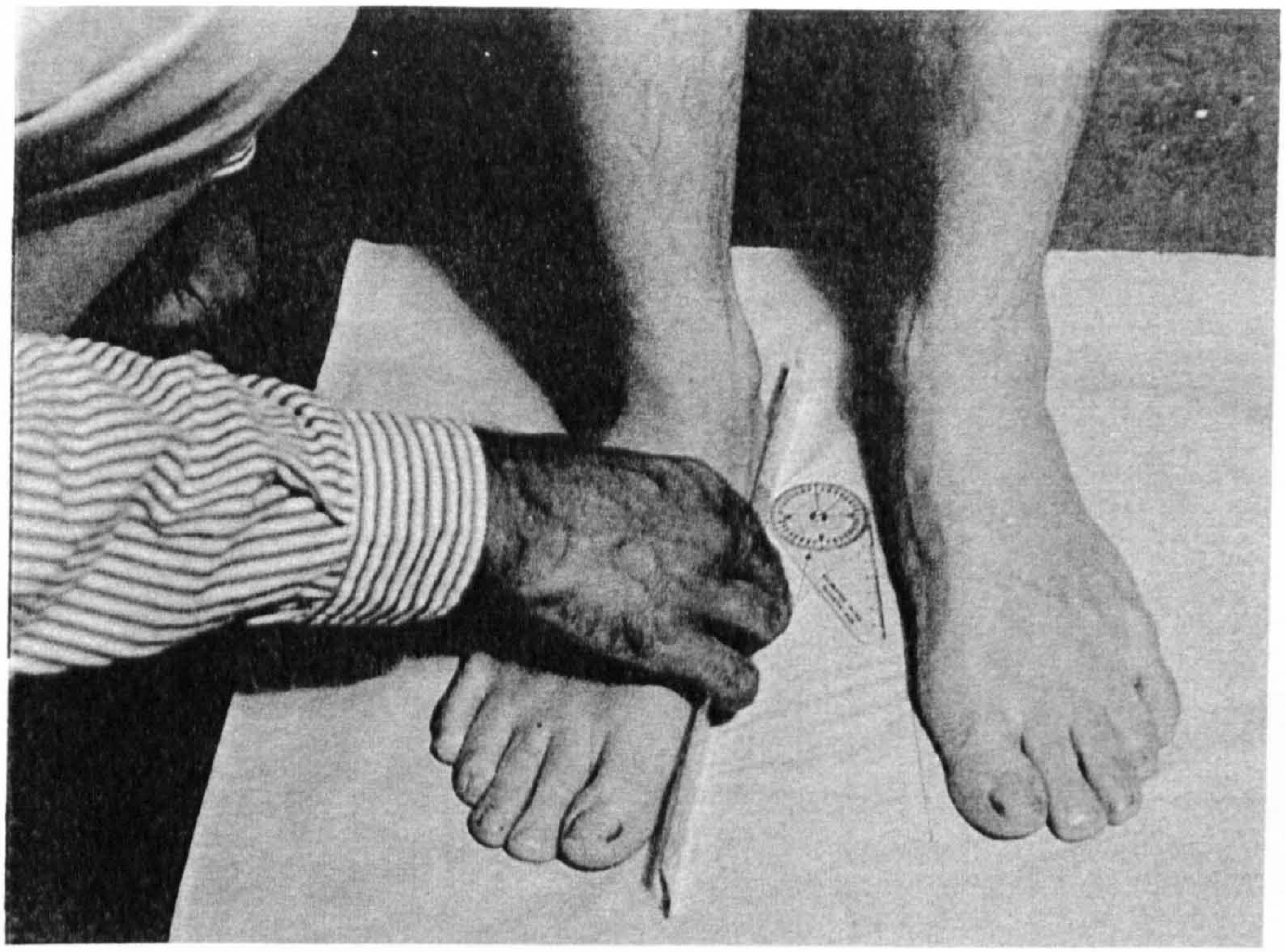

Figure 2.11. Method of measuring the angle of gait.

there is a rearfoot varus; if, however, it is less than 90^0 then there is a rearfoot valgus. The angle between the midline of the leg and the horizontal determines whether there is a *tibial varum* or *tibial valgum* (figure 2.12b). Again, this angle is normally 90^0. If it is greater than 90^0, as measured on the medial side of the leg, then there is a tibial varum (bow leg); but if it is less than 90^0 then there is a tibial valgum (knock knee). Quite often tibial varum is accompanied by rearfoot varus; and a tibial valgum is accompanied by a rearfoot valgus (**Neale, 1981**). The former combination is associated with a narrow base of gait, while the latter is associated with a wide base of gait.

Ranges of motion can be measured at other joints in the foot as part of the biomechanical examination. However, for a basic appraisal of foot function, the above assessment can provide a lot of useful information. As an example, it may be possible to predict the types and locations of future mechanically induced pathologies. With a forefoot varus, the soft tissue beneath the fourth and fifth metatarsal heads is likely to develop a thick area of callus or lesion because of excessive weight-bearing on the lateral side of the foot during locomotion (**Weed and Holt**[1]).

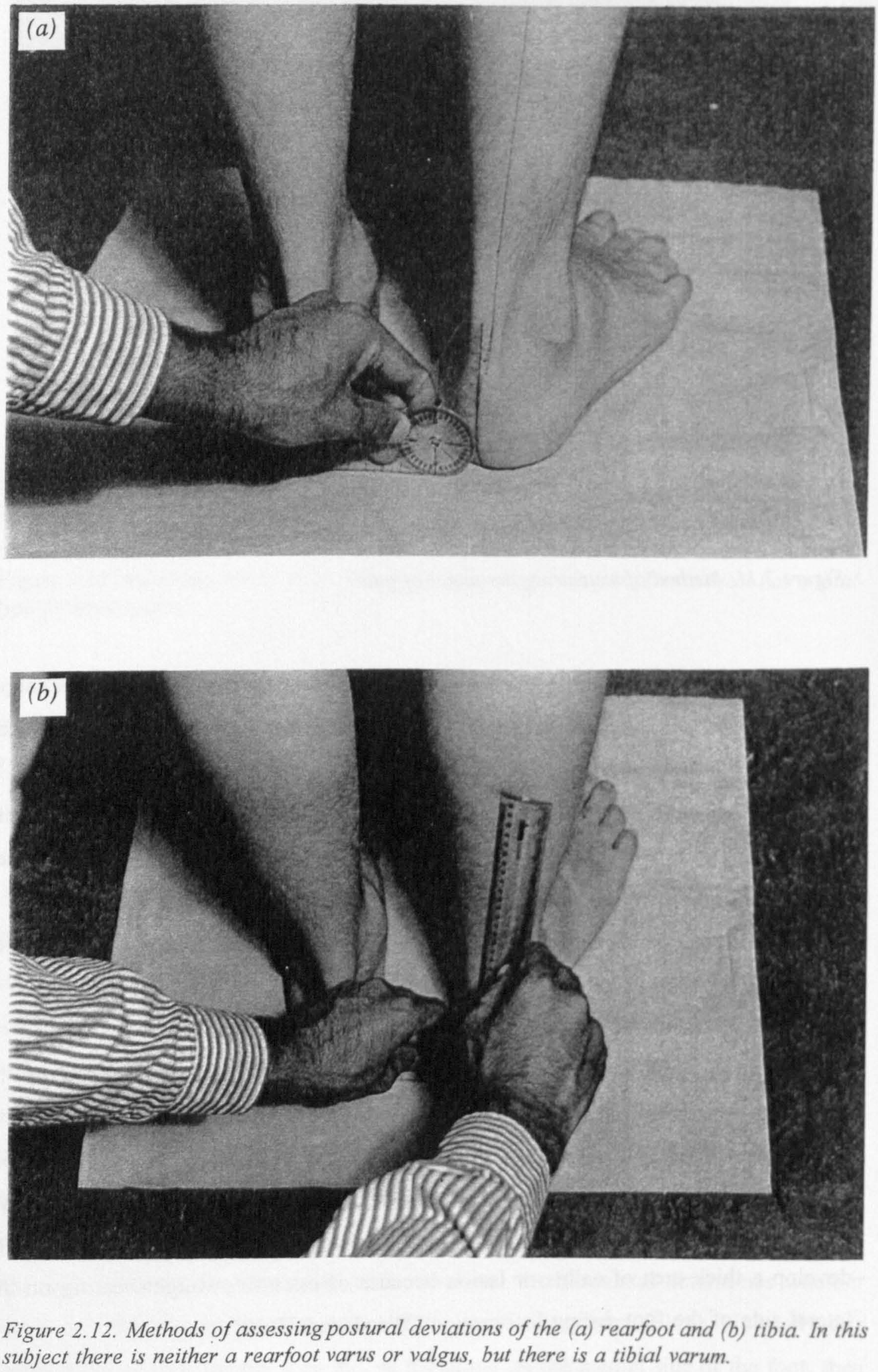

Figure 2.12. Methods of assessing postural deviations of the (a) rearfoot and (b) tibia. In this subject there is neither a rearfoot varus or valgus, but there is a tibial varum.

One of the most important functions of the foot is to assist in locomotion. Although the term locomotion encompasses both walking *and* running, only walking will be discussed in this section. During walking at least one foot is in contact with the ground at all times. This is in contrast to running, during which there is a period when both feet are off the ground.

2.3.1 normal gait

Walking can be broken down and analysed as a series of repeating events. The successive occurrence of one of these events defines a *gait cycle*. Any event can be taken as the starting point of a gait cycle, but it is usual that this event be *heel-contact*, ie. the initial contact of one of the heels with the ground.

Four *spatial* terms are used to describe the relative positions of the feet on the ground during a gait cycle (figure 2.13). The *stride length* is the distance covered from heel-contact to heel-contact of the same foot; the *step length* is the distance covered from heel-contact of one foot to heel-contact of the other; the *stride width*, or *base of gait*, is the distance between the centre points of the heels; and the angle between the feet is known as the *angle of gait* - most people "toe-out" during walking.

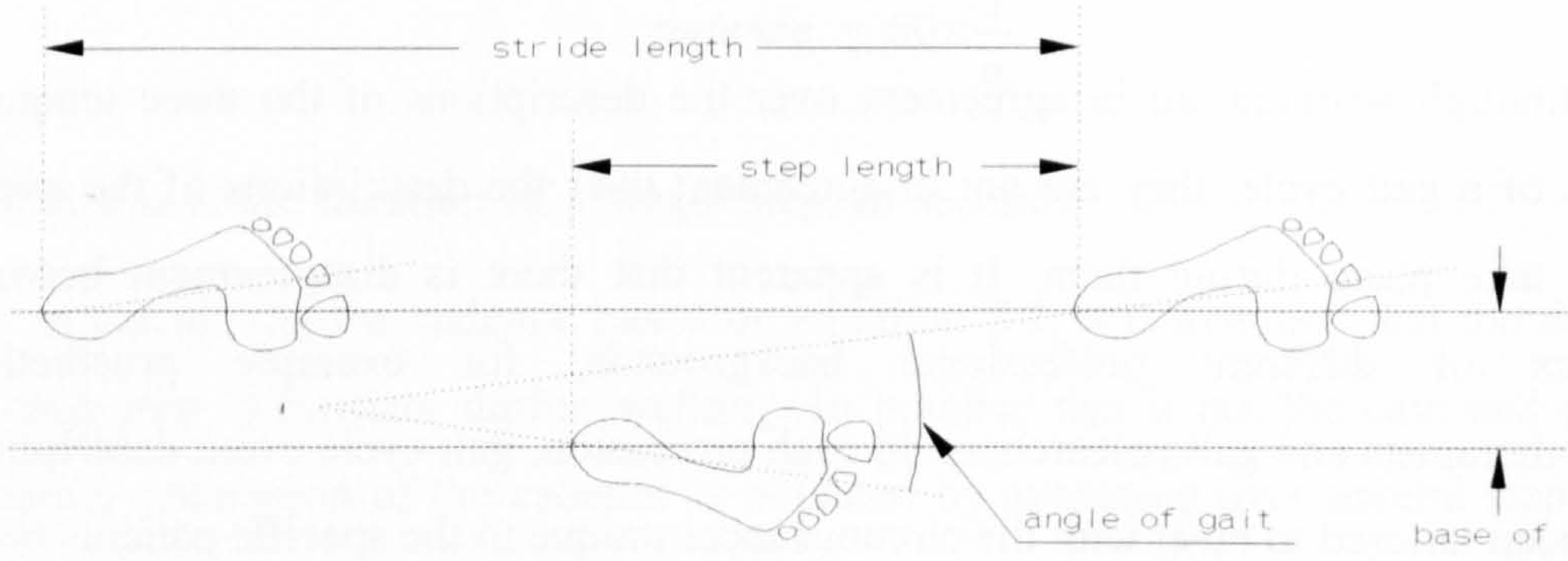

Figure 2.13. Spatial terms used to describe the relative placement of the feet on the ground during a gait cycle.

During walking, each foot is alternately on the ground and swinging forward through the air. These *temporal phases* of gait are respectively known as *stance* and *swing*. In addition there is a brief period of *double-support* when both feet are on the ground (figure 2.14).

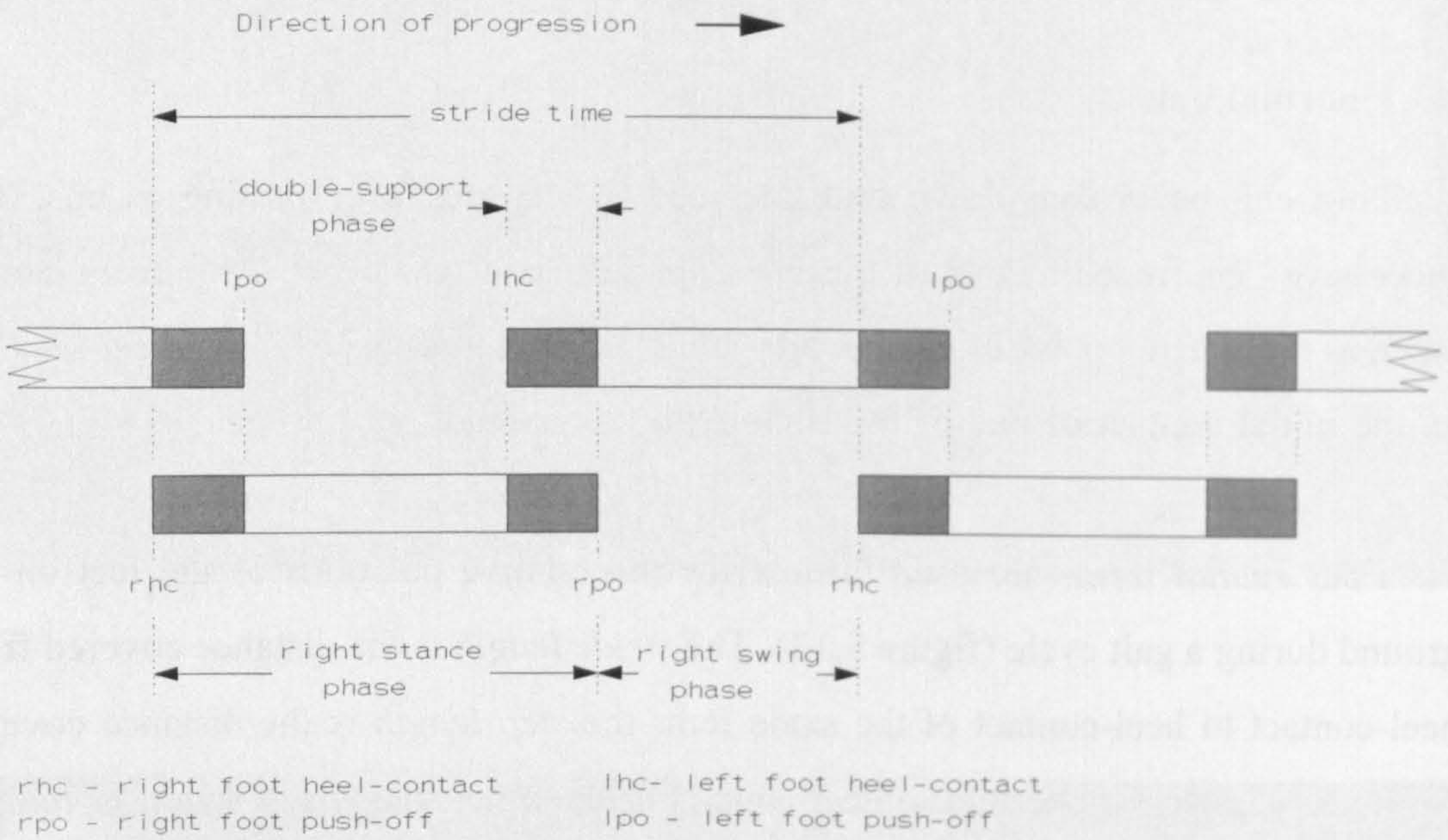

Figure 2.14. The three temporal phases of gait: stance, swing and double-support. The periods of double-support are shaded.

Although workers are in agreement over the descriptions of the three temporal phases of a gait cycle, they are not in agreement over the descriptions of the events which take place during them. It is apparent that there is disagreement between workers of different professional backgrounds, for example prosthetists, physiotherapists and gait researchers. In each profession, gait cycle event descriptions have been tailored to meet with the circumstances unique to the specific patients being studied. **Thomas and Supan (1990)** provide summaries of the different gait cycle event descriptions that are currently being used in a number of professions. The description of gait cycle events given here is taken from **Mann (1991)**, a doctor of medicine involved in gait research.

The stance phase is considered to last for about 62% of the gait cycle and to consist of five events: *heel-contact*, *foot-flat*, *midstance*, *heel-rise* (or *heel-lift*) and *push-off*. The swing phase lasts for the remaining 38% of the cycle and consists of three events: *acceleration*, *midswing* and *deceleration*.

Beginning with heel-contact, the foot typically reaches a flat position as early as 7% into the cycle. The period between foot-flat and midstance, when the leg is vertical, lasts for the next 27%. At 34% into the cycle, the heel begins to leave the ground; and at 50% the heel of the opposite foot makes ground contact. During this latter stage, both feet are on the ground in a period of double support, which lasts for about 12% before the first foot leaves the ground to go into its swing phase. The swing phase begins with a period of acceleration, when the foot initially leaves the ground and continues forward through the air. Midswing is a momentary period, which corresponds to midstance of the opposite foot. Finally, during the period of deceleration, the foot is being slowed down in preparation for the heel to contact the ground to begin another cycle.

The number of *steps* taken in a minute is known as the *cadence*. The simplest method of calculating the cadence is to determine the duration of a single step and then to use the following equation:

$$cadence = 60 \times \frac{1}{\alpha} \qquad \textbf{(2.1)}$$

where α is the duration of a single step, in seconds.

In calculating the cadence based on equation **2.1**, it is assumed that the duration of each step is constant during walking. In practice this is not the case and a more accurate calculation of the cadence is obtained by averaging over several steps.

movements of the foot and leg during gait

Gait cycle events are normally accompanied by specific movements of the foot, leg and the rest of the body. Movements of the foot and leg during walking are described below referencing the work of **Neale (1981)** and **Bevans (1992)** throughout. However, because of the complexity of upper body movements, these are not described here.

Instead, for a description of upper body movements during walking, see other sources such as **Whittle (1991)**.

Prior to heel-contact. In preparation for the postero-lateral border of the heel to contact the ground first, the foot is slightly dorsiflexed and inverted as it nears the end of its swing phase. Dorsiflexion of the foot occurs about the ankle joint; and inversion about two joints: the subtalar joint and longitudinal axis of the midtarsal joint. Just prior to heel-contact, the heel of the opposite foot lifts off the ground.

Heel-contact to foot-flat. During this first period of the stance phase, the foot is used to slow the body down and to steady it. At heel-contact, the leg internally rotates, and the foot rapidly plantarflexes and everts to bring the forefoot onto the ground to provide stability. Because the foot is inverted at heel-contact, the lateral side of the foot contacts the ground first and weight is then gradually transferred medially across the forefoot.

Foot-flat to midstance. Early in this period, the lower leg reverses its motion and begins to externally rotate. This causes the foot to evert at the subtalar joint, which increases the range of motion available at the midtarsal joint. The foot becomes mobile for a greater part of this period and able to adapt to uneven terrain. However, at the end of this period, the joints of the foot will be aligned in their neutral position, with the lower leg vertical. In neutral, the midtarsal joint is maximally everted, which makes the foot a more stable base of support.

Midstance to heel-lift. As the body moves forward over the supporting leg, the foot and toes dorsiflex. With continuing forward movement, the gastrocnemius muscle tenses and the heel is *pulled* off the ground. As the heel lifts, the foot plantarflexes and inverts.

Heel-lift to push-off. During this final period of the stance phase, the foot becomes a rigid lever in preparation for propelling the body forward. The foot continues to plantarflex and invert, but the weight supported by the foot decreases as there is a gradual transfer of weight in a lateral-to-medial direction to the opposite foot. As more weight is borne by the toes, they dorsiflex further at the metatarsophalangeal joints. This causes the plantar aponeurosis to tense, which elevates the longitudinal arch through the "windlass" mechanism. In this way a rigid lever is created out of the foot, which is used to provide forward leverage for the body. Push-off during level walking typically occurs about the transverse metatarsophalangeal joint axis (**Bojsen-Moller and Lamoreux, 1979**) (figure 2.2), with the hallux parallel to the direction of progression. As the foot leaves the ground the plantar aponeurosis relaxes.

The movement of the metatarsals during forefoot loading is of special interest in this thesis. Upon forefoot loading the metatarsals splay as the foot everts (**Shereff et al, 1990**). This increased forefoot width enables the foot to maintain stability. The forefoot is also able to adapt to uneven terrain during loading, which is made possible by the independent mobility of the most medial and lateral metatarsals. The first metatarsal has ample downward and outward mobility, while the fourth and fifth have ample downward and inward mobility (**Viladot, 1991**). The second and third metatarsals are the least mobile because of proximal ligamentous ties and the interaction of their bases with the tarsal bones, and are subsequently the most stable metatarsals. As the heel lifts off the ground, the metatarsals incline more to the horizontal as they rotate about their heads. Between foot-flat and push-off, the first metatarsal head rotates *in situ*; while the lesser four rotate and simultaneously translate forward (**Price, 1959**, see figure 7.8). The firm contact of the big toe with the ground prevents the first metatarsal head and the plantar skin beneath it from slipping forward. In the case of the lateral four metatarsals, as they incline, their heads rotate and translate forward *with* the underlying pad of tissues (*op cit*). The tips of the four corresponding toes provide a gradual braking to this roll, which can be verified by the measurement of pressure underneath them (**Hughes et al, 1990**).

Swing phase. During the swing phase the foot must be clear off the ground. To facilitate this, during the first half of the swing phase the foot dorsiflexes and everts. However, during the second half of the swing phase, when the foot is preparing to contact the ground again, it inverts slightly.

2.3.2 pathological gait

Even to an untrained observer, several characteristics of a normal gait can be easily detected. For instance, there should be distinct heel-contact and push-off events; symmetrical movements of both legs and feet; repeated alternate advancement of one leg in front of the other; and swinging of the arms close to the sides of the body out of phase with the swinging of the legs. In an abnormal gait, there is a clear departure from these, and other, *normal* gait characteristics (**Whittle, 1991**). Abnormalities of gait may result from disorders affecting the nervous system, muscles or joints. For example, patients with Parkinson's disease (a disease affecting the brain) have a characteristic shuffling gait (**Murray et al, 1978**). However, an abnormal gait does not only result from disease. For instance, a subject physically able to walk *normally*, but who has just collided with an object, may adjust their gait to avoid pain occurring in the feet or another part of the body.

The fixed forefoot and rearfoot positions of forefoot varus, forefoot valgus, rearfoot varus and rearfoot valgus described earlier, may also cause specific gait abnormalities. These gait abnormalities are mainly confined to the movements of the foot and are in themselves difficult to recognize when looking at a patient walking. However, the fixed forefoot and rearfoot positions may also lead to the development of specific plantar pathologies which can be identified. This section describes the compensatory motions of the foot during walking in the presence of forefoot varus, forefoot valgus, rearfoot varus and rearfoot valgus; and describes associated plantar pathologies that may arise due to the abnormal loading of some parts of the foot. The work of **Weed and Holt**[1] and **Neale et al (1981)** is referenced throughout.

forefoot varus

In this condition the fifth metatarsal head is lower than the first in relation to the rearfoot when the foot is in the neutral position. A forefoot varus is fully compensated when the forefoot is completely contacting the ground during midstance. This is accomplished by eversion at the subtalar and/or midtarsal joints. If, however, the forefoot remains inverted at midstance, due to a limited range of evertory motion at the subtalar and/or midtarsal joints, the condition will have been only partially compensated.

A *fully compensated forefoot varus* results in a classical flat-foot, ie. a foot which has abnormal eversion on loading resulting in a low medial longitudinal arch and very little motion in gait. During the stance phase, as the heel lifts off the ground the foot will invert slightly at the subtalar joint, but not back to the neutral position. This results in an "unlocked" forefoot on rearfoot, ie. the forefoot is not fully everted about the midtarsal joint. As a result, the first ray[2] is made unstable by becoming *hypermobile*. Under these conditions, push-off can only occur after the other foot has accepted much of bodyweight. In the foot with a fully compensated forefoot varus, callus may develop beneath the second metatarsal head as a consequence of the hypermobile first metatarsal. When the first metatarsal bears weight it will tend to dorsiflex excessively, resulting in increased loading of the adjacent second metatarsal head.

In the foot with a *partially or uncompensated forefoot varus*, callus may develop beneath the fourth and/or fifth metatarsal heads. These calluses may arise because of the frontal plane deviation of the forefoot in combination with inadequate compensatory motion, causing abnormally prolonged loading on the lateral side of the forefoot.

forefoot valgus

In this condition the first metatarsal head is lower than the fifth in relation to the rearfoot when the foot is in the neutral position. A forefoot valgus is again fully compensated when the forefoot is completely contacting the ground during midstance. This is accomplished by inversion at the subtalar and/or midtarsal joints. However, momentarily after this happens, and just before the heel lifts off the ground, the subtalar and midtarsal joints evert so that load can be transferred medially across the forefoot.

In the foot with a forefoot valgus, callus may develop beneath the first and/or second and/or fourth and/or fifth metatarsal heads. Calluses on the medial side of the

---

[2] A *ray* is formed by a metatarsal and the phalanges anterior to it.

forefoot may arise because of the frontal plane deviation of the forefoot causing abnormal loading; whereas calluses on the lateral side of the forefoot may arise because of the abnormal compensatory motion that may take place in gait.

rearfoot varus

In this condition the rearfoot is inverted in relation to the lower leg when the foot is in the neutral position. Compensation for a rearfoot varus takes the form of abnormal eversion at the subtalar joint, which begins as the forefoot bears load and continues until the calcaneus is perpendicular. During this period there is also abnormal eversion at the midtarsal joint as greater load is taken by the forefoot, which continues until heel-lift. At push-off, the foot will invert slightly at the subtalar joint, but not back to the neutral position. Inversion will also take place at the midtarsal joint during this period.

In the foot with a rearfoot varus, callus may develop beneath the fourth and/or fifth metatarsal heads because of biased loading. However, if the fifth metatarsal can dorsiflex enough on loading, callus may only develop beneath the fourth metatarsal head.

rearfoot valgus

In this condition the rearfoot is everted in relation to the lower leg when the foot is in the neutral position. With this condition the foot is everted at the subtalar joint *throughout* the stance phase. Consequently, for the lateral side of the forefoot to contact the ground, inversion is forced to take place at the midtarsal joint during midstance. This is abnormal because the forefoot should be everted during this phase. However, as a result of this motion the forefoot is made hypermobile. Rearfoot valgus is associated with classical "flat-foot", and combined with a hypermobile forefoot, gait becomes apropulsive with very little force on the digits at push-off.

In the foot with a rearfoot valgus, callus may develop beneath the second metatarsal head because of abnormal shearing in this region during forefoot loading.

2.4 THE FOOT AND THE SHOE

More often than not, daily activity takes place in some form of footwear. From an early age, shoes and socks are typically worn which provide both a comfortable environment for the foot while at the same time protecting it from injury. In a recent review article, however, **Staheli (1991)** cited several studies which suggested that shoes may in fact impair the natural development of the foot, and stated that optimum foot development occurred barefoot. In particular, the wearing of shoes has been shown to cause progressive narrowing of the forefoot (**Hoffman, 1905**) and to be linked with hallux valgus deformity (a lateral deviation of the big toe) (**Sim-Fook and Hodgson, 1958**). These forefoot problems may occur because the shoe style chosen may cause cramping of the toes throughout foot development and in later life (**Bonney and Macnab, 1952**); or the shoe may have been incorrectly fitted and tight in the forefoot. The shoe tends to restrict movement at the joints in the foot and foot function, although the extent to which this occurs depends on the shoe construction and fit. Physical restrictions placed on the foot can, however, be tolerated as long as the shoe still provides a comfortable environment and a normal gait is still possible.

A number of factors define a "good" shoe style and a "good" shoe fit which may result in the prevention of forefoot deformities; give better overall comfort; and allow greater range of movement at the joints. These factors are discussed in this section.

Although there are seven basic shoe styles (**Rossi, 1985**), only the characteristics of the lace-up shoe are discussed here. Lace-up shoes, or shoes with a fastening over the instep, are usually supplied to patients requiring corrective or accommodative shoes because they possess more of the features which define a "good" shoe. The lace-up shoe holds the foot securely with the fastening over the instep, tends to be wide in the forefoot and have a low heel.

2.4.1 shoe structure

The lace-up shoe consists of an *upper*, a *sole* and a *heel* (figure 2.15). The upper consists of two main parts: the *quarter*, which surrounds the heel and the instep; and the *vamp*, which covers the dorsum of the midfoot, forefoot and the toe area. Situated beneath the lacing is the *tongue*, a strip of material which prevents the lacing from irritating the dorsum of the foot.

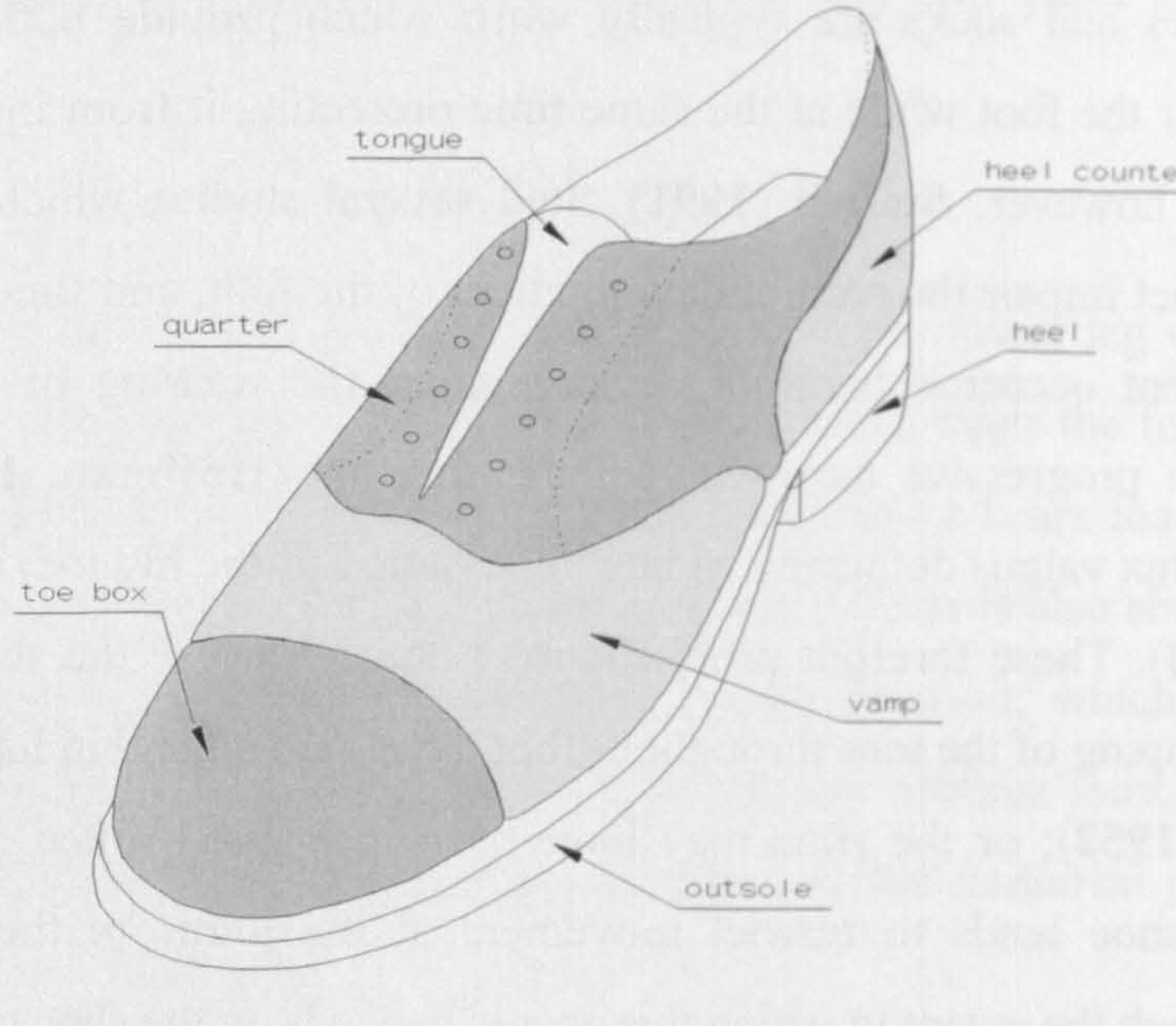

Figure 2.15. Anatomy of a Gibson lace-up style shoe.

Shoe uppers have traditionally been made of leather, a "breathable" material which absorbs perspiration and moulds to the shape of the foot during wear. More recently, synthetic materials have been used in the making of shoes, which although not performing as well as leather, are cheaper and lighter. To maintain the shape of a shoe made from these soft materials, stiffeners are used behind the heel and over the toes in areas respectively known as the *heel counter* and *toe box*.

The upper is not made to the exact shape as the foot. Practically this is not possible because the foot changes shape during walking and normal function would be impaired by such a foot covering. Instead the upper is a modified shape of the foot and is smooth in contrast to the foot's irregular shape. Shoes are made with reference to specific foot dimensions with *allowances* added or taken away. For instance, allowances are usually added to the length of the shoe to give room for the toes to extend during weight-bearing and walking; and are usually taken away from the girth in order to ensure the finished shoe is tight over the instep to prevent the foot from slipping forward.

The sole of the shoe usually consists of an inner lining (*inner sole*, *insock* or *sock liner*) of leather; an outer sole of either leather or a synthetic material; and a compressible filler in between. A *shank* made of metal, wood or a synthetic material is inserted between the inner and outer soles to prevent excessive flexing of the sole in the midfoot area. The shank runs from the heel to the widest point in the forepart of the shoe - the *tread line*. At push-off, the forefoot dorsiflexes at the metatarsophalangeal joints and the sole of the shoe should also flex in this region. Some shoes, however, have rigid soles which restrict this dorsiflexion (**Bojsen-Moller and Lamoreux, 1979**). In order to compensate for this the whole foot must tip forward during push-off, which requires the sole to be elevated at the toe end. In general, the more rigid the sole of the shoe, the greater the toe elevation (*toe spring*) needs to be to enable normal gait.

2.4.2 shoe fit

The fit of a shoe is largely subjective and governed by such factors as the style of shoe; the hosiery being worn with the shoe; the preference for a tight or loose fitting; and the time of day the shoe is tried on - the foot can swell by up to 10% during the course of the day (**Diebschlag, 1978**, cited in **Snijders, 1987**). However, although largely subjective there are a few sensible *guidelines* concerning a "good" shoe fit that when adhered to may reduce discomfort during walking and prevent foot pathologies.

A "good" fitting shoe should: (1) be tight around the heel, while still being comfortable; (2) be tight over the instep to hold the foot firmly back and prevent forward sliding and cramping of the toes in the toe box; (3) have a toe box that is both deep and wide enough to accommodate the toes without dorsal and lateral pressure; (4) have an allowance of 10-12 mm between the end of the shoe and the longest toe in order to prevent the toes from jamming when they extend during walking (**Janisse, 1992**); (5) accommodate the first metatarsophalangeal joint in the widest part of the shoe; and (6) when laced, leave a gap of about half-an-inch in the quarter to allow for further adjustment.

For most people shoe selection is by trial fitting: several shoe sizes are tried-on before one is chosen. In some shoe shops foot length and ball width (the distance across the widest part of the forefoot) are measured before a particular shoe size is selected for fitting. However, it is possible to find two feet with identical length and ball width measurements that are unable to fit comfortably into a single size shoe. This is because the feet may be physically different in other respects, for instance the line of the metatarsal heads (the metatarsophalangeal break line) may be at different levels so the feet may have different longitudinal arch lengths (**Hughes, 1991**); or the feet may adopt different shapes and sizes during activity. Fortunately, with the vast number of shoe designs available in a range of sizes *and* width combinations, off-the-shelf shoes can be found to fit most feet *comfortably*. However, there are the exceptions of particular patients that may require specially made (*bespoke*) shoes. In providing bespoke shoes to patients with foot problems the guidelines above should still be followed, but it is particularly important to ensure the foot is held back in the shoe with the use of laces or a velcro strap across the vamp; and that pressure is not applied to the sides of the forefoot, or to any dorsal deformities, by the upper. Contoured insoles may also be needed by some patients to accommodate prominent plantar deformities or pathologies.

2.5 DYNAMIC PLANTAR STRESS MEASUREMENT

2.5.1 introduction

The measurement of stresses between the soles of the feet and a supporting surface during walking is a specialised form of gait analysis that can be used to study foot function and assess plantar pathologies in particular patient groups.

There are two approaches to the measurement of plantar stresses, namely registration between the barefoot and ground or between the plantar surface and shoe insole. Devices to measure the pressure distribution between the barefoot and ground were the first to be developed. In 1882 **Beely** (cited in **Elftman, 1934**) stood subjects on a bag filled with rapidly setting plaster of Paris and interpreted the deepest impression left in the negative cast produced as the area of the foot carrying the greatest load. This interpretation was not wholly correct since a deep impression could also be due to an abnormality of the foot shape. In fact the information obtained by this method predominantly relates to the foot shape rather than pressure and as a consequence quantitatively useful pressure measurements cannot be derived.

Since this early method, devices for measuring stresses between the barefoot and ground have continued to be developed. However, in the western world where the wearing of shoes as opposed to going barefoot is more common, a variety of professional groups would find measurements made between the foot and shoe insole more useful. These would include shoe manufacturers, orthopaedic surgeons, orthotic manufacturers, chiropodists, clinicians, and those conducting biomechanics research. In the past, a lack of suitable instrumentation has been the limiting factor in the development of in-shoe devices. Recently, with greater innovations in electronics, more emphasis has been placed on the development of these devices and in reducing the obtrusiveness associated with in-shoe data collection.

Pressure and shear constitute the components of stress that may be measured between the foot and its supporting surface. To date most barefoot and in-shoe devices have been limited to measuring pressure. The reason for this is twofold: firstly, there has been a lack of suitable instrumentation for the development of devices to measure shear; and secondly, pressure has been thought to be greater in magnitude than shear stress and so potentially the more damaging component. In 1963 **Bauman et al** proposed shear to be just as damaging as pressure to the plantar soft tissues through a tearing action, but it is only in the last fifteen years that devices have emerged to quantify shear stresses and determine their pathogenic role, if any, in the development of blisters and plantar ulcers.

An historical account of the many devices developed for measuring the stress distribution between the barefoot and ground, and between the foot and shoe insole during walking is presented later in this section. This is followed by a discussion of clinical findings from plantar stress studies conducted with asymptomatic subjects. To begin with, the terminology that will be used is defined and the technical problems associated with plantar stress measurement are discussed.

2.5.2 terminology

Surveying the literature of plantar stress studies reveals the interchangeable use of the engineering terms, force, stress and pressure; and the use of the descriptive terms *local peak pressure*; *global peak pressure*; and *maximum peak pressure* without clear meaning. All of these terms are clarified here in order to lay the basis for precise usage in the remainder of this thesis.

force, stress and pressure

Force is a quantity that tends to change the relative shape or motion of the body on which it acts. From the work of Sir Isaac Newton (1642-1727), the motion of a body was found to be constant unless it was acted upon by an external force; and the subsequent acceleration of the body was found to be proportional to the force applied. This theory of motion is known as *Newton's second law*, or *the law of inertia*, and can be written in the following equation form:

$$F = \frac{dmv}{dt} \tag{2.2}$$

where F represents force, m mass, v velocity and t time.

or, for constant mass,

$$F = m \times a \tag{2.3}$$

where F represents force, m mass, and a acceleration.

From équation 2.2, the Systéme International (SI) unit of force, the *Newton* (symbol *N*), is defined as the force required to give a unit mass of 1 kg a unit acceleration of 1 ms^{-2}.

Force is a *vector* quantity, which means that it is specified by both its magnitude and direction. As such, when discussing forces, the direction in which they are acting should be stated unless this is obvious. It is customary to specify three mutually perpendicular components of force based on the Cartesian coordinate system. The *direct*, or *normal force*, acts perpendicularly to a surface. If the surface is horizontal then normal forces are also vertical. Acting tangentially to a surface are two mutually orthogonal components of *shear force*. In this case, if the surface is perpendicular to the vertical, then the tangential forces are also horizontal.

Stress is directly related to force and is defined as the limit of force over the area on which it acts as the area tends towards zero,

$$\sigma = \lim_{da \to 0} \frac{df}{da} \tag{2.4}$$

where σ represents stress, f force, and a the area over which the force acts.

In practice, stress is approximated by being defined as force distribution, ie. the intensity of a force acting over a unit area,

$$\sigma = \frac{f}{a} \tag{2.5}$$

The SI unit of stress is the *Pascal* (symbol *Pa*), which is defined as the uniform distribution of 1 Newton over an area of 1 m^2. Stress, like force, is a vector quantity and like force it is also customary to specify three mutually perpendicular components.

Pressure is a sub-set of stress. In fact pressure is a *normal stress* and as such has the same SI units as stress. The terms pressure and normal stress can therefore be used interchangeably in the context of plantar stress studies.

local peak, global peak and maximum peak

When describing values of plantar pressure or stress recorded by a device, clinical significance is placed upon the highest value of pressure or stress recorded. In this context the words "peak" and "maximum" are often used. In this thesis, strict definitions of these terms are adopted with respect to spatial and temporal factors, respectively, to provide clarity.

When looking at a snapshot of the pressure distribution beneath the foot at one instant in time (spatial analysis) there may be *peaks* of pressure evident. When pressure is recorded at discrete locations beneath the foot against a timebase (temporal analysis) there will be a *maximum* value of pressure registered at each location.

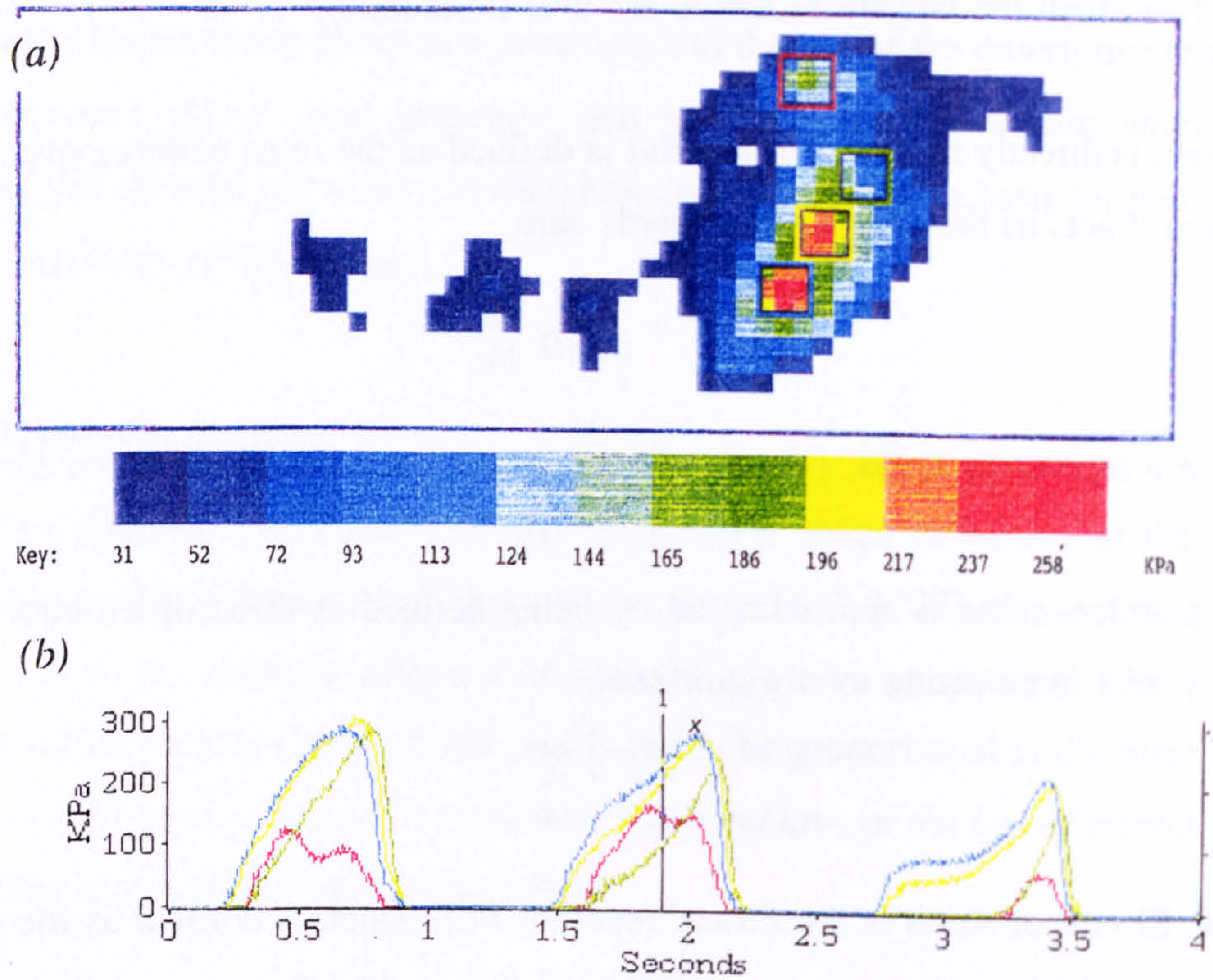

Figure 2.16. Spatial and temporal terms used to describe the pressure peaks generated beneath the foot during walking: (a) local peak, global peak and (b) maximum peak pressures.

Local peak. Looking at the pressure distribution beneath the foot at a particular instant in time there may be several discrete areas of high pressure (figures 2.16a and 6.3). The magnitudes of pressure in these areas at this instant are *local peak* pressure values.

Global peak. The *global peak* is the one local peak that is greater than all the others in magnitude at any particular instant (see the area bounded by the blue box in figure 2.16a).

Maximum peak. When looking at the site of a local peak over time, the term *maximum peak* describes the *highest* value observed in the time period (see the peak 'x' on the graph in figure 2.16b).

2.5.3 technical problems in plantar stress measurement

The values of plantar stress measured by a device will be affected by the presence of the device itself and/or the device limitations which will limit its accuracy. As a result, two different devices may give different values of peak pressure beneath the metatarsal heads of one subject, which will depend on the particular physical characteristics and/or performance characteristics of the device being used. In the following discussion, various terms used when describing the characteristics of a device are defined in order to provide an understanding of how a device itself may influence the data it measures and to what extent.

perturbation

It is important that the transducer making the measurement does not disturb the true situation. This may seem obvious, but it is a problem that is very often overlooked. The early devices for barefoot plantar pressure measurement included a surface that deformed considerably when loaded. In this way the device itself provided some shock absorption, but more importantly it increased the area of the plantar surface in contact with the support surface thereby reducing peaks of pressure compared to walking on a hard surface.

Several workers using discrete transducers have tended to tape them directly to the skin for both barefoot and in-shoe measurements. The interposition of a rigid transducer between the foot and a supporting surface has the effect of introducing a local hard spot. As a result, inaccuracies may be introduced into data from four possible sources: (1) the subject may alter their gait in order to reduce the pain caused by walking on the transducer; (2) stresses may be concentrated at the site of the transducer, with the result that pressure peaks and shear stresses may be increased; (3) the transducer may be imprecisely positioned beneath the sites of interest; or (4) the transducer may move during walking due to shear. For in-shoe measurements the ideal solution would be to have a paper-thin sensor shaped to fit inside the shoe (an *insole sensor*) and which the subject could not feel.

spatial resolution

A large number of very small, closely spaced sensors affords a device a high spatial resolution. With such a device it is possible to discriminate between pressure peaks beneath closely spaced anatomical sites, for example the metatarsals heads. A device with large sensors cannot do this and in addition gives lower values for peak pressures due to spatial averaging.

A device with a very high spatial resolution is necessary in order to measure the *true* maximum peak pressures. In a theoretical analysis, **Lord (1993)** concluded that *true* maximum peak pressures could only be measured by averaging peaks of pressure over an area of 0.1 mm x 0.1 mm. At present there is no system with a resolution this high for in-shoe measurements and so all studies of in-shoe plantar pressure unintentionally underestimate the *true* maximum peaks.

measurement range

An obvious point to remember when designing a measuring device is that it should be able to measure the *expected* magnitudes of stress. An optical device (*pedobarograph*) used by **Boulton et al (1983)** was apparently not able to record pressures above approximately 1.16 MPa (11.8 $kgcm^{-2}$). This is obvious from the pressure-time graphs presented, which have *saturated* at this value (*op cit*).

load resolution

A device should be able to discriminate increments of stress. Ideally these increments should be very small, for example 5 kPa or maybe smaller, to give the device a high load resolution and increased accuracy in measurement. The load resolution of a device is expressed as a percentage of its measurement range. Printing devices for pressure measurement have very poor load resolution and also tend to saturate at very low pressures, ie. show blotting for all pressures above a certain value. One in particular, the *Harris and Beath Footprinting Mat* only allows four ranges of pressure to be clearly distinguished (0-3 kPa, 3-12 kPa, 12-26 kPa and 26-47 kPa), with pressures above approximately 47 kPa (4.7 $kgcm^{-2}$) being beyond its measurement range **(Silvino**

et al, 1980). It is possible to design a printing device with a higher measurement range, for example, as high as 700 kPa to measure the normal ranges of pressure generated during walking (figure 2.17). However, if such a device had a load resolution of say 10%, ie. 70 kPa, a vast amount of detail would be lost since pressures of approximately 30-70 kPa are generated beneath a large area of the foot during walking (see figure 5.3).

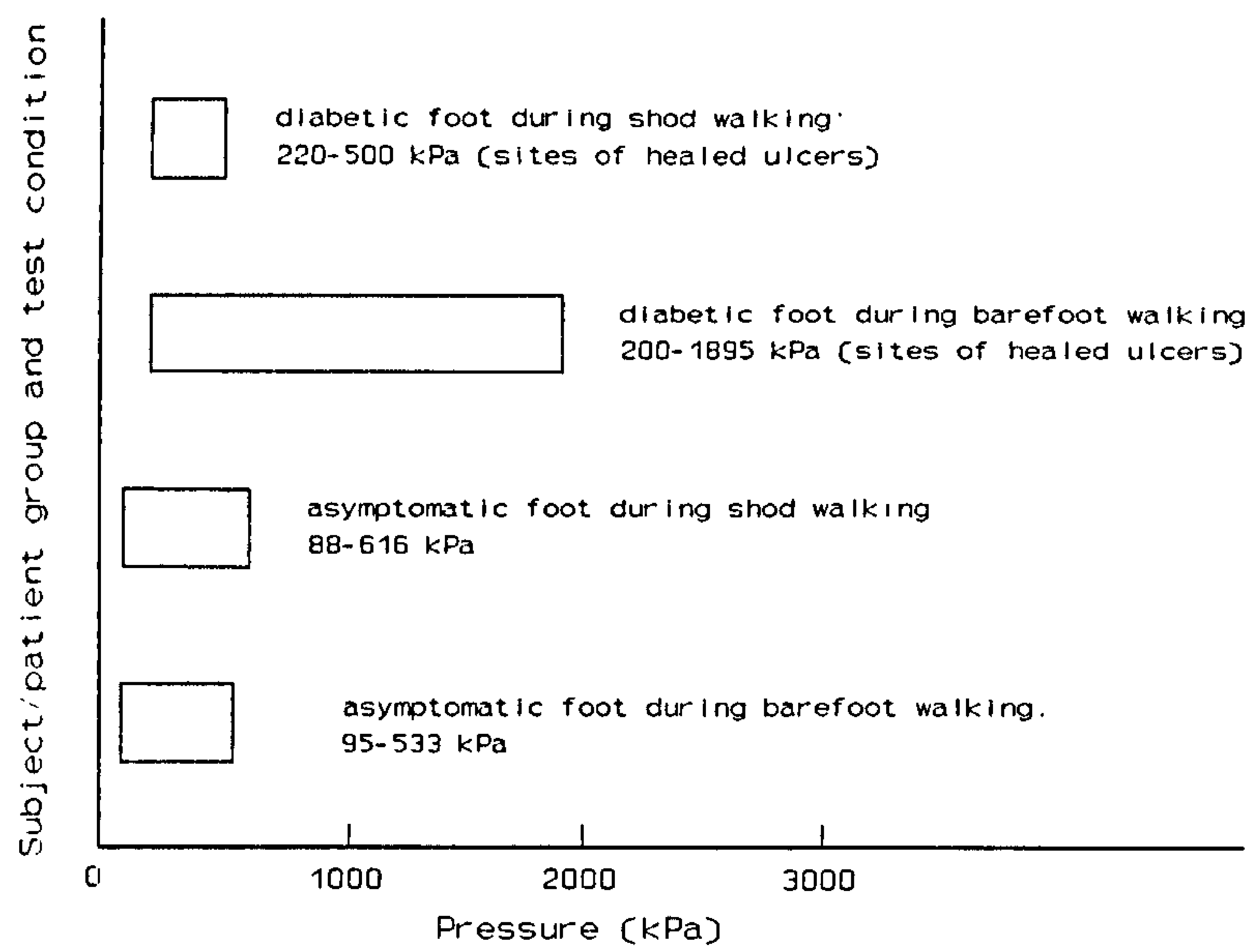

Figure 2.17. Ranges of average maximum peak pressure measured by workers beneath the asymptomatic and diabetic forefoot (see tables 2.3 and 2.4, and section 2.6.4).

calibration

Calibration is performed on a device in order to determine its response to load. Most workers calibrate their device statically, ie. by placing weights onto it and then removing them. However, when using a device to record dynamic plantar stresses it is more appropriate to determine its response under similar, dynamic conditions. Dynamic calibration can be achieved by applying a step, ramp or sine wave input load to the device. A step input load takes the form of an abrupt change from one steady value to another; a ramped input load varies linearly with time; and a sine wave input load is of a cyclic nature. Dynamic calibration is ideally performed by applying a sine wave input load, the frequency of which should extend up to the expected highest frequency content during walking - 75 Hz according to **Simon et al (1981).**

linearity

For a given change in applied load, the output voltage signal should change in proportion, regardless of where in the measurement range of the device that change is. That is, if the load changes from 10 N to 15 N and the output voltage changes by 0.5 volts (V), then for *any* 5 N change in load, the output should change by 0.5 V. If a device shows a linear response to load and no offset, ie. no output for no load, then a conversion factor (a single number) can be used to convert all subsequently recorded data to pressure or shear stress. Unfortunately, most devices are non-linear. This being the case there are two approaches to converting device output to stress. Either the calibration data is incorporated into software to convert individual voltages into stress; or a best fit straight line is made to the non-linear calibration data, the gradient of which becomes the data conversion factor. Using this latter approach, it is important to have an idea of the linearity error of the device in order to determine the degree of inaccuracy incorporated into the data. Linearity error is calculated by fitting a first order polynomial (a best fit straight line) to a graph of the calibration data and calculating the maximum voltage difference between the line and the graph for an applied load value. Linearity is an expression of this difference as a percentage of the full scale voltage output.

hysteresis

Hysteresis is an indication of the different response of a device during loading and unloading. As an example, if an additional 50 N load is applied to a device loaded with 100 N and the output changes from 2 V to 3 V, then when the 50 N load is removed the output should return to 2 V. A device exhibits hysteresis if the output measured before a load is applied is different from the output measured when this load is applied and then removed. When such data is plotted it results in a characteristic looped calibration graph (see figure 3.5). From this graph, hysteresis is calculated as the maximum voltage difference between the loading and unloading parts of the graph for an applied load value, expressed as a percentage of full scale voltage output.

frequency response

As a device is loaded at higher and higher frequencies, there is a gradual deterioration in its output response. Either there is a time delay between the input and the output, or the magnitude of the output does not change in proportion to the input, or both. Such a device has a poor frequency response and may not make accurate measurements when subjected to loads with a high frequency content. Because the highest frequency component during walking is approximately 75 Hz (**Simon et al, 1981**), a measuring device with a frequency response of 75 Hz or higher is ideal.

sampling rate

The rate at which data is sampled as a device is loaded is determined by the cut-off frequency of the filter used in the interface electronics. Electronic filters are used to stop unwanted signals of certain frequencies being included in the recorded data. The highest frequency component generated when walking across a forceplate is approximately 75 Hz (**Simon et al, 1981**). As a result, the interface electronics between a forceplate and recording system may include a filter with a 75 Hz cut-off frequency. With a higher cut-off frequency it is possible that high frequency noise signals from various sources may corrupt the data. A lower cut-off frequency may, on the other hand, result in the loss of important gait data. If a filter is included in the interface electronics with a cut-off frequency at 75 Hz, then in accordance with the Nyquist sampling theory, the data sampling rate should be at least twice this, ie. 150 Hz. In practice this only provides two sample points per cycle, so the sampling rate is more often between three and ten times the maximum frequency of interest. The higher rates of sampling would be optimal, but tend to be restricted by the available computer RAM (*Random Access Memory*).

temperature response

Certain materials used in constructing a measuring device may make it sensitive to temperature changes. Changes in temperature may therefore change the output, regardless of whether the device is being loaded at the time or not. To determine the response of a device to temperature, it is usually left *unloaded* and then heated over a range typically never exceeding 80^0C. Temperature sensitivity is a measure of the voltage output per unit rise in temperature ($V/^0C$). More often, the voltage output is converted to load using calibration data and the temperature sensitivity is then quoted as the apparent applied load per unit temperature rise ($N/^0C$).

Temperature sensitivity is more of a problem when making in-shoe measurements because the measuring device is typically subjected to temperature fluctuations: plantar skin temperature is somewhere in the region of 5^0C to 10^0C below core body temperature (**Cavanagh and Ulbrecht, 1991**). Most workers try to minimise the effect of temperature changes on device output by allowing the device to equilibrate for a period in the shoe before taking measurements. However, temperature fluctuations may still occur during walking due to air circulation brought about by the "pumping action" of the foot in the shoe (**Wilson and Peet, 1994**).

2.5.4 methods for barefoot plantar pressure measurement

All the devices developed for measuring barefoot plantar pressures are floor-mounted and measure less than three-quarters of a metre square. As a result of this there are two major disadvantages with these devices concerning the amount and the quality of data that is collected. Firstly, only one step can be analysed out of a walk consisting of several; and secondly, gait may be contrived in order to walk onto the device.

The earliest devices in this category were mechanical and shared the flaw of including a surface that deformed *considerably* when the barefoot made contact. This undoubtedly affected the observed pressure distribution and the magnitudes of local pressure recorded. With advancements in signal processing and electronic hardware, barefoot devices have been developed overcoming this problem.

a review of techniques

Devices for measuring pressure beneath the barefoot and ground during walking have been extensively reviewed elsewhere (**Lord, 1981; Lord et al, 1986; Alexander et al, 1990**). Only a brief review of these devices is given here.

A number of workers have developed devices based on ink printing. The first was **Frostell** in 1925 (cited in **Horan, 1976**) who suspended a wire frame over an inked pad and stood subjects on a piece of paper laid over the top. With this device, areas of high pressure resulted in a much greater transfer of ink to the paper, ie. the ink in areas of high pressure was darker and spread over a larger area compared to areas of low pressure. Following on from this, several workers developed printing devices based on the deformation of rubber matting. **Morton (1930)** was the first of these with his *kinetograph*. The rubber mat, which had a flat upper surface and deformable ridges on the underside, was laid on top of an inked fabric which covered a sheet of paper. Walking across the mat produced a print of parallel lines with widths proportional to the locally applied pressure. The print resolution of this technique was improved by **Harris and Beath (1947)** who modified the matting to a grid pattern in which the ridges were at three different levels. The highest ridges compressed under relatively low pressures, and the lower ridges compressed under progressively higher pressures. The *Harris and Beath Footprinting Pad*[3] is still in clinical use today and perhaps endures because it can be used to assess plantar problems quickly and at low cost. However, it is unfortunate that both of these devices have poor load resolution and produce blotting at very low pressures giving them limited measurement ranges. More recently, **Grieve (1980)** replaced ink printing in favour of indenting aluminium foil using a rubber mat with pyramidal projections. This method, although more expensive than printing, does provide a much clearer record, which is also more easily quantifiable.

Direct visualisation devices enable the instantaneous plantar pressure distribution to be directly observed and filmed. The first device developed to do this worked on a similar principle to **Morton's** kinetograph, except that loading deformed the pyramidal projections on the underside of a rubber mat against a glass plate **(Elftman, 1934)**. This process was viewed and filmed from below, with the image enhanced by the introduction of a white opaque fluid between the mat and glass. A variation on this basic design was described by **Arcan and Brull (1976)** in a device that was initially developed for static measurements, but was later developed commercially for dynamic

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[3] Downs Surgical (Escalab) Ltd. Park Way Close, Park Way Industrial Estate, Sheffield S9 4WJ, England.

measurements[4]. The original device, which utilised the principle of optical interference, included a glass plate which was overlaid with a sandwich consisting of an optically sensitive sheet, a reflective layer and a polarising layer. This was then loaded via a "flexible" sheet consisting of a matrix of widely spaced protrusions, which was necessary to "discretise" the load. Loading produced circular interference fringes, the diameter of which were a calibratable function of force. The commercial development of this device consisted of an array of five hundred hemispherical ended rods, which were used to load a glass plate coated with a photoelastic material (**Simkin and Stokes, 1982**). The interference fringes were filmed with a cine-camera at forty-eight frames per second. Unfortunately the spacing of the rods again gave this device a poor spatial resolution. In the latest photoelastic method to be developed, the spatial resolution was considerably improved by using a flexible plastic loading device with grooves spaced 3 mm apart that were divided every 3 mm along (**Rhodes et al, 1988**). The drawback with this device was the time consuming nature of the calibration procedure. Another optical principle was utilised by **Chodera** (cited in **Lord, 1981**) in a device that provided a spatially continuous indication of static plantar pressure. The original *pedobarograph*, which was the forerunner of a dynamic version, consisted of an edge-lit glass plate that was covered with a plastic sheet. When loaded, the contact between the plastic and glass destroyed the condition for total internal reflection and escaping light illuminated the underside of the plastic. The intensity of illumination was proportional to the applied pressure. In a number of more recent publications, **Betts et al (1980a,b,c,d)**, **Duckworth et al (1982)** and **Franks et al (1983)** described the development of a dynamic pedobarograph. The pressure image produced by the subject walking across this device was reflected in a mirror and filmed from the side at a rate of twenty-five frames per second. The dynamic pedobarograph is now commercially available[5] and has a high frequency response of 12.5 Hz, which is over the 10 Hz considered adequate for gait studies (**Betts et al, 1980a; Lord and Chodera, 1980**).

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4 The 'Footprint', marketed by Ramot Ltd, Tel Aviv, Israel; and later by Palrod Scientific Instruments, Ein-Harod, Israel 18965.

5 The "Dynamic Pedobarograph" marketed by Biokinetics, 6 Wortley Moor Road, Leeds, LS12 4JF, UK.

Although lacking a spatial resolution as high as the pedobarograph, pressure mat devices are thinner and lighter, and as a result are more portable. The Nicol mat was the first electrical pressure mat system to be developed. This consisted of a rubber mat with 16 conductive strips on each side. The strips, which were orthogonal, formed 256 capacitive transducers, the capacitance of which varied in proportion to the applied pressure. Each transducer was sampled at 200 Hz. All pressure mat devices are very similar in design, but differ in their electrical operating principle. These differences are noted in two tables below. Table 2.1 gives details of the *Nicol mat* (**Nicol and Hennig, 1978**) and two commercial devices that were developed from it; while table 2.2 gives details of other pressure mat systems. The device developed by **James et al (1982)**, consisting of an array of conductive rubber transducers, was marketed by WM Automation[6] as the *Musgrave Footprint*. This device was further improved by the Company, who developed a mat with an active area of 193 mm x 412 mm incorporating two-thousand-and-forty-eight transducers. The latest development of the Musgrave Footprint, which is now marketed by Preston Communications Limited[7] (**Van Der Meer and Weir, 1988**), employs force-sensitive resistive technology (FSR). Loadcells created using FSR technology basically consist of two conductive materials that form a resistive path when pressed together: the more pressure applied, the lower the resistance. TekScan Inc.[8] also employ FSR technology in their *Bigmat* system, but have produced a much thinner mat with two-thousand-and-sixty-four, 8 mm x 8 mm pressure-sensitive cells. A device developed by **Cavanagh and Hennig (1982)** used small squares (5 mm x 5 mm) of a piezoelectric material, *lead zirconate titanate* (PZT), which generate an electrical charge proportional to an applied force. Silver electrodes were diffusion bonded to the major surfaces of the squares and five-hundred-and-twelve of these were laid onto a printed circuit board. Each element was connected to a common ground and the assembled device was then attached to a rigid substrate and covered with a heat insulating material.

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6 WM Automation, 76 Valley Duff Drive, Carnmoney, Newtownabbey, Co. Antrim, BT36 6PB, N Ireland.

7 Preston Communications Ltd, New Ross, Dinbren Road, Llangollen, Clwyd LL20 9TF, UK.

8 TekScan Inc, 4th Floor, 451 D Street, Boston, MA. 02210, USA.

Table 2.1. Summary of pressure mat devices developed from the Nicol mat.

Source/Company	Device name	Mat dimensions	Number of transducers / type / dimensions
Nicol & Hennig (1978)	Nicol mat	240 mm x 480 mm	256 / capacitive / 13 mm x 20 mm
Novel$_{gmbh}$[9]	EMED System	Range: 100 mm x 100 mm to 310 mm x 480 mm	2 to 9 per cm^2 / capacitive / NA
Clinical Interactive Research[10]	Electropodograph		1024 / capacitive / NA

NA - Not Available

Table 2.2. Summary of pressure mats other than the Nicol mat.

Source/Company	Mat dimensions	Number of transducers / type / dimensions
James et al (1982)	406 x 203 x 25 mm^3	512 / electrically resistive carbon loaded elastomer / NA
Cavanagh and Hennig (1982)	150 x 375 mm^2	512 / piezoelectric / 5 mm x 5 mm
Musgrave Footprint (Van Der Meer and Weir, 1988)	630 x 325 x 32 mm^3	2048 / force-sensitive resistors / NA
TekScan Inc.[8]	500 x 500 x 2 mm^3	2064 / force-sensitive resistors / 8 mm x 8 mm

NA - Not Available

2.5.5 methods for in-shoe plantar stress measurement

In-shoe plantar stress measurement has two advantages over barefoot measurements. Firstly, the important interface between the plantar surface of the foot and shoe insole can be investigated; and secondly, data from several consecutive steps can be recorded and analysed.

Each one of the barefoot devices described above enabled the entire plantar surface to be investigated. In contrast, the early systems developed for in-shoe measurement used small numbers of discrete transducers. This required the investigator to have some

[9] Novel$_{gmbh}$, Beichstrasse 8, 8000 Munchen 40, Germany.

[10] Clinical Interactive Research, Avenue des Fauvettes, 2A Grasmussenlaan - 1950 Kraainem, Belgium.

knowledge of foot anatomy and of various foot pathologies to ensure that sites of importance were not overlooked. Most often, the plantar aspects of the 5 metatarsal heads, the heel and the hallux were the chosen sites for measurement, and the transducers were stuck directly to the skin before the shoe was put on. It could reasonably be assumed that fixing transducers to the foot in this way would have an adverse effect on the data being recorded. In particular, the thickness and rigidity of the transducer would act as a local hard spot and cause pain to the subject with the result that gait may be altered; and may also concentrate stresses at the site of the transducer. To avoid these problems, some workers have instrumented inlays, ie. flush-mounted discrete transducers into custom-made insoles. However, **Bransby-Zachary et al (1990)** maintain the data recorded in this way is not a true representation because the inlay may alter impact loads. Recently, thin pressure sensitive insoles with very high sensor resolution have been developed, utilising methods of transduction employed in the pressure mat systems described above. These insole systems overcome the problems of accurately placing discrete transducers and the effects of thick, cushioning inlays. Several of these have been marketed and now appear in clinical use.

In the following discussion of transducers for in-shoe plantar stress measurement, transducers have been classified according to the principle of transduction.

capacitive transducers

A capacitive transducer basically consists of two conducting plates separated by an insulating layer. The insulating layer tends to be elastic, so that applying pressure to the plates decreases their separation thereby increasing the electrical capacitance. Electronic hardware is then used to interface the transducer to a chart recorder or computer for data display.

The very first disc-like transducer to be used for in-shoe pressure measurement was capacitive. Developed by **Schwartz and Heath** and reported in 1947, six of these 12.7 mm diameter, 2.286 mm thick devices were taped beneath each foot for measurements beneath the hallux, first, third and fifth metatarsal heads, and the medial and lateral aspects of the heel. The transducers were connected by a twenty-five foot flexible cable to a mechanical data recorder. Inside the recorder, each transducer was electrically connected to a galvanometer, which reflected light to expose a moving roll of photographic paper. Before use, a static calibration was performed to ensure each device had a reproducible output over the pressure range 0 to 241 kPa (0 to 35 psi) to within 5% of a standard calibration curve. A smaller capacitive transducer (11.3 mm diameter and 1 mm thick) was described by **Bauman and Brand (1963)**. Only five of these devices were used in tests, taped beneath the hallux, first, second and fifth metatarsal heads and the centre of the heel. Non-linearity exhibited above 350 kPa (3.5 $kgcm^{-2}$) in static calibration was compensated by non-linear amplification. This extended their linear range to 500 kPa (5 $kgcm^{-2}$).

A larger transducer was described by **Holden and Muncey (1953)** to measure in-shoe heel pressures. This heel-shaped device was 3.175 mm thick and had a large load sensitive area of 322.5 mm^2 which unfortunately greatly averaged the pressures recorded. Anticipating very high frequency components during gait, the transducer was designed to have a frequency response from zero to 2000 Hz, but only a static calibration was performed before use.

Two capacitive insole systems are commercially available for direct pressure measurement. Infotronic[11] market the *Computer Dyno Graph System* (CDG System), which provides special slippers incorporating eight prepositioned 30 x 30 x 1.5 mm^3 transducers. A data sampling rate of 50 Hz per transducer is possible, over a period of 20 seconds. Data is initially stored in a unit worn by the patient and later downloaded to a computer for analysis. The transducer has a limited accuracy of 10% full scale output. The *EMED System* marketed by $\text{Novel}_{\text{gmbh}}$[9] employs transducers similar to those developed for the Nicol mat (**Nicol and Hennig, 1978**). This is an insole system which supplies insoles in two standard sizes, 6 and 8, although other sizes can be made to order. The size 8 insole accommodates seventy-two transducers, one per 2 cm^2. Each transducer is individually calibrated over the range 20 kPa to 1.5 MPa (2-150 Ncm^{-2}) and claimed to be insensitive to shear. The data sampling frequency is selectable from the range 20-200 Hz. This top frequency limit is much

[11] Infotronic Medical Engineering, PO Box 73, 7650 AA Tubbergen, Netherlands.

higher than the sampling frequency of the CDG System and is more appropriate for gait studies.

strain gauge transducers

Many conductive materials undergo a change in electrical resistance when subjected to mechanical strain. A strain gauge transducer utilises this phenomenon. The strain gauge itself is usually no more than an etched track on a thin circuit board. This is usually bonded to a metal beam, which deflects in proportion to the applied load. The strain gauge is wired into a Wheatstone bridge circuit to convert the small changes in electrical resistance to voltage. Three types of strain gauge are available: the wire gauge, the semiconductor gauge and the foil gauge. Most workers have used the semiconductor types in their transducers.

Three groups of workers have employed a cantilever beam principle in their transducer design. The first was **Lereim and Serck-Hanssen (1973)** who produced a 12 mm diameter, 2.5 mm thick transducer, which was mounted into a PVC inlay of similar thickness. The transducer consisted of a membrane on a mounting ring, which was turned from a single piece of brass. Pressure applied to the membrane deflected an underlying silicone beam, which underwent a change in electrical resistance. Weight-bearing X-rays of the feet were used as a guide to position a total of five of these transducers into an inlay. The transducers were either placed beneath the metatarsal heads or beneath the hallux, heel, first and fifth metatarsal heads, and the base of the fifth metatarsal. **Soames et al (1982)** produced a much thinner transducer from beryllium copper. This 13 x 13 x 0.9 mm^3 device had a 3 mm x 5 mm pressure sensitive area, which was a cantilever beam to which a semiconductor strain gauge was bonded. To obtain accurate measurements there were a number of exacting requirements: the load had to be evenly distributed on the cantilever beam; the plantar soft tissues had to be compliant to deflect the recessed beam; and the surface beneath the transducer had to be flat and rigid so that the transducer did not deform and obliterate the recess into which the beam deflected. Fifteen sites were chosen for measurements with the transducers taped to the foot. These sites were beneath each metatarsal head and toe, the base of the fifth metatarsal, and the medial, lateral and posterior aspects of the heel.

Similar to **Lereim and Serck-Hanssen (1973), Frost and Cass (1981)** mounted their transducers into an inlay, this time made of rubber. The transducer consisted of a strain gauge sandwiched between rubber discs. Six of these 6 mm diameter, 1.6 mm thick transducers were mounted into the inlay to be beneath the first, fourth and fifth metatarsal heads, the hallux, apex of the third toe, and the centre of the heel. As the transducer was loaded, flattening of the rubber produced horizontal extensile Poisson strains, which were detected by the strain gauge. Unfortunately, there were significant problems with this device. The rubber used was a non-linear material, which exhibited considerable hysteresis and creep; data was corrupted by bending deformation of the transducer; and the wires leading to the transducer frequently broke when a shoe containing the instrumented inlay was put on. In the end, because of this latter problem, the inlay was taped to the foot for measurements.

In a study of plantar pressure in different styles of shoe, **Bransby-Zachary et al (1990)** used a commercially manufactured transducer taped beneath the third metatarsal head. The transducer was a sealed unit, 3.15 mm in diameter and 0.9 mm thick, containing a silicon strain gauge bonded to a metal diaphragm. These workers reported the device to have excellent performance characteristics: hysteresis and nonlinearity produced a combined full-scale error of 0.5%; the device was temperature stable; and had a flat frequency response up to 100 kHz. However, quantising error is noticeable on the pressure-time graphs presented, suggesting low bit resolution of the analogue-to-digital (A/D) converter being used and a source of measurement inaccuracy. At the time of their report, these workers were in the process of developing a multi-transducer system.

force-sensitive resistive film

A number of recently developed commercial systems employ force-sensitive resistive (FSR) technology to produce very thin and in some cases very high resolution sensors for placing inside the shoe. Two designs of FSR sensor can be described. In one design, plastic sheets are used (usually Mylar®) onto which a conductive polymer is deposited in tracks. The tracks run along and across individual halves of the insole and when these are glued together a pressure sensitive cell is created where perpendicular tracks cross. In an alternative design, metal electrodes are printed onto a lower plastic sheet and a conductive polymer or elastomeric sheet is laid on top. As pressure is applied, the tracks or the electrodes and the conductive material make contact and a resistive path is created. Electrical resistance decreases as the pressure applied increases.

The French company Midi-Capteurs[12] produce an insole-shaped transducer with one-hundred-and-twenty-seven sensors. A flexible Kapton printed circuit of one-hundred-and-twenty-seven electrodes is overlaid with a layer of conductive rubber to produce an insole 2.5 mm thick. **Peruchon et al (1989)** conducted a study of in-shoe plantar pressures using this insole and described in detail its characteristics and the interfacing electronics they developed for it. Calibration of the sensor was performed by applying a ramped force at 1 Hz, which showed the sensor to exhibit considerable non-linearity and hysteresis due to the rubber in its construction. In subject tests, each transducer was sampled at 40 Hz and together with 128 kbits of computer RAM a walking sequence of 25 seconds duration could be recorded. It was found that the useful lifetime of the insole was limited to one hundred experiments due to mechanical abrasion causing deterioration of the rubber layer.

Three American companies currently market foot pressure measurement systems based on FSR technology. In 1982 the Langer Biomechanics Group[13] introduced their *Electrodynogram System* (EDG System). Seven flexible sensors are provided for each foot, which can be individually placed and are stuck to the plantar surface. The sensors have outer dimensions of 15.9 x 12.7 x 0.39 mm^3, with a smaller pressure sensitive area, 11.1 x 5.5 mm^2. A waist-mounted unit provides signal conditioning and stores data for later downloading to a computer for analysis. Data is sampled at 100 Hz for 5 seconds per sensor **(Polchaninoff, 1983)**. Interlink Electronics[14] manufacture their

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[12] Midi-Capteurs International Systems, 4 Rue des Libellules -31400 Toulouse, France.

[13] The Langer Biomechanics Group UK, Unit 7, The Green, Cheadle, Stoke-On-Trent, England.

[14] Interlink Electronics, 535 E.Montecito Street, Santa Barbara, CA 93103, USA.

FSR sensors to customer requirements. As a result, their sensors find use in many applications other than foot pressure measurement. For most applications, Ultem® polyetherimide film is used for depositing the conductive materials. A typical insole may be 0.25 mm thick, with sensors 9 mm in diameter spaced 13 mm between centres transversely and 19 mm between centres longitudinally. In Company literature, a sensor lifetime of 10 million+ actuations is claimed for forces applied in the range 0.1 N to 200 N. However, it is made clear that the sensors are not suited for exact force measurement. The TekScan[8] *F-Scan Gait Analysis System* provides an even thinner insole sensor, just 0.1 mm (see figure 4.1). The sensor consists of a Mylar substrate onto which 3.5 mm wide conductive polymer tracks are deposited. The tracks run along and across individual halves of the insole and when these are glued together a pressure sensitive cell is created where perpendicular tracks cross. The cells are spaced 5 mm between centres. In the first version of the system (version 1.22), nine-hundred-and sixty loadcells were created by this method on the standard US size 14 insole supplied (UK 12, Continental 47). In the latest version (version 3.611) there are nine-hundred-and-fifty loadcells on the standard insole. To date, this system provides the highest resolution sensor for in-shoe measurements. However, the novelty is in the flexibility of the insoles and the track architecture, which allows the insoles to be trimmed with scissors from a US size 14 down to a minimum US size 3 (UK size 2, Continental 35) before shoe insertion. An ankle-mounted unit provides signal amplification and computer-sited electronics perform signal analogue-to-digital conversion. The system is capable of interfacing two sensors at a time to allow data to be sampled from both feet simultaneously, with the user able to select data sampling rates between 1 and 100 Hz per cell.

piezoelectric transducers

Piezoelectric transducers utilise natural crystal materials such as quartz or manufactured crystal materials such as lead zirconate titanate, which generate an electrical charge proportional to an applied force. In contrast to capacitance, strain gauge and FSR transducers, no external voltage supply is needed. Instead the crystals are connected via electrodes to amplifiers which collect the output charge and convert it to a voltage. However, in electrically connecting an amplifier or any measuring instrument, there is a tendency for the crystal to discharge, which results in a transducer with a poor steady state response. This problem is overcome by using amplifiers with very high input impedances (*charge amplifiers*), which unfortunately are very expensive and add considerably to the cost of these systems.

A synthetic barium titanate crystal was used by **Hennacy and Gunther (1975)** in their transducer. Connections to the crystal were made by soldering low capacitance cables to two opposite silvered surfaces and the crystal was then coated in epoxy. After hardening, the surfaces were sanded flat to produce a device with a pressure sensitive area of 735 mm^2. Both static and dynamic calibrations were performed. However, the low frequencies used in dynamic calibration, between 0.33 Hz and 1.16 Hz, do not justify the validity of this device for accurate registration during gait, which can contain frequency components as high as 75 Hz (**Simon et al, 1981**). Six sites were chosen for measurements, with the transducers stuck to the skin with spray adhesive and tape. Data was recorded from beneath the first, third and fifth metatarsal heads, and the medial and lateral aspects of the heel. The sixth transducer was placed over the navicular prominence. The temperature sensitivity of the transducers was taken into account by allowing them to equilibrate for 10 minutes on the foot. In-shoe tests were subsequently conducted on a treadmill. This system was further developed by **Gross and Bunch (1988)** who improved the transducer design and the signal conditioning electronics. The transducers were modified by first soldering copper tabs to either side of a lead zirconate titanate (PZT) piezoelectric crystal before soldering tin wire to the tabs. This was done to alleviate the possibility of uneven loading across the crystal caused by soldering a wire directly to it. The crystal was then enclosed by vacuum forming polyurethane over one side and sticking urethane tape over the other. The exact dimensions of the assembled transducer were not given. Calibration was performed by applying impulse stresses of approximately 2 MPa for times of up to 200 ms and using a Kistler[15] piezoelectric force transducer as a reference. A total of eight transducers were made and calibrated and between them displayed less than 3.4% linearity and less than 5.8% hysteresis. In subject tests, transducers were stuck to the under surface of an insole (thickness not given) that was taped to the foot. This

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[15] Kistler Instrument Corporation, 75 John Glenn Drive, Amherst, New York 14120, USA.

method of mounting provided a flat surface for attaching the transducers and minimised point loading when the foot was in the shoe. However, this methodology would undoubtedly redistribute plantar pressure, leading to an underestimate of true peaks of pressure. Transducers were located beneath the hallux, the first, second, third and fifth metatarsal heads, the base of the fifth metatarsal, the centre of the heel, and the inferior aspect of the navicular prominence. A waist unit (6 x 11 x 3 cm^3) housed the charge amplifiers, which were stated to provide a simplified and more effective means of noise shielding and d.c. offset control than the electronics used by **Hennacy and Gunther**. One limitation of this system was its inability to resolve static loading patterns. Dynamic tests were therefore conducted with the subject running on a treadmill at 3.58 ms^{-1} and the signals sampled at 333 Hz per transducer.

A greater number of smaller transducers were employed by **Hennig et al (1982)** in their insole system. The transducers were constructed from lead zirconate titanate crystals to which silver electrodes were diffusion bonded. Four-hundred-and-ninety-nine of these transducers (4.78 mm^2 and 1.2 mm thick) were laid on a thin copper gauze cut to the shape of a US size 10 insole (UK size 8, Continental 41), which served as a common ground. Individual copper wires were then soldered to the top of each crystal and a layer of silicone rubber was laid on top. The finished insole was between 3 mm and 4 mm thick, with the transducers spaced 6 mm between centres. Dynamic calibration was performed on each transducer after the insole was made. Peak stresses as high as 1500 kPa were applied, but very low frequencies were used - between 1.43 Hz and 5.88 Hz. Linearity and hysteresis were found to be less than 2% and 1%, respectively; while investigation of the temperature sensitivity showed this to be less than 1.5% between 10 ^{0}C and 40 ^{0}C. The charge amplifiers were contained in a backpack (25 x 18 x 15 cm^3, weighing 2.9 kg), which was connected via cable to a PDP-11/34 computer. The entire insole could be scanned at up to two-hundred frames per second, but in subject tests data was sampled at half this rate.

Since the early 1980's, a number of research groups have developed systems utilising polymers which possess piezoelectric qualities. **Pedotti et al (1984)** used PolyVinylidene Fluoride (PVdF) film just 200 μm thick to produce a UK size 8 insole (US size 10, Continental 41) with 16 sensors. Circular aluminium discs, 6 mm in diameter, were deposited onto the film and served as electrodes for each sensor. It was noted that PVdf was temperature sensitive, which could make the calibrated output signal unreliable. For this reason, these workers measured the ratio between the instantaneous value of force on each sensor and the absolute maximum value of force measured by any of the 16 sensors over time, rather than measuring absolute pressure values. A larger pressure sensor was produced by **Bhat et al (1989)** using Kynar PVdF film, 50μm thick, sandwiched between Mylar insulating layers and taped to a metal backing plate. Seven of these, 42 x 19 x 2 mm^3, sensors were embedded in an inlay, but at the time of publication this system had yet to be used for plantar pressure measurements. Plantar pressures recorded will be greatly averaged though because of the sensors size. A group working at the University of Kent have recently become active in PVdF transducer research. Using PVdF and copolymer films, **Nevill (1991)** produced a direct pressure transducer measuring 10 x 10 x 2.8 mm^3. The piezoelectric film was sandwiched between a brass layer and a double-sided circuit board. Coax cable was soldered to the brass terminal and through a groove cut into the circuit board another coax lead was potted with an epoxy adhesive against the piezoelectric film. Only a static calibration was performed, which showed both linearity and hysteresis to be within 1.5% for maximum loads of 20 kg and 10 kg, respectively. Two inlays were made with eight of these transducers mounted in each, beneath the hallux, metatarsal heads, lateral arch, and the centre of the heel. At present, electronics are being designed at Dundee University to interface thirty-two of these transducers, which will eventually be placed into a single inlay (personal communication). In addition, a piezoelectric transducer for shear measurement along one direction, ie. *uni-axial* measurement, has been developed at the University of Kent, which is only slightly thicker than their pressure transducer (10 x 10 x 3.4 mm^3) (**Akhlaghi and Pepper, 1993**).

magneto-resistors for shear stress measurement

With the exception of the last principle of transduction described, transducers have been designed around the others for pressure measurement only. In this section, a principle of transduction is introduced which has been exclusively used in transducers designed to measure shear stresses.

The magneto-resistive principle of transduction relies on the resistance of a semiconductor magneto-resistor varying as the strength of a surrounding magnetic field varies. The magneto-resistor is wired as two arms of a Wheatstone bridge circuit and electrically excited. With a magnet located centrally above the element the bridge arrangement is balanced. If the magnet is displaced, the bridge circuit becomes unbalanced producing an electrical signal proportional to the magnet displacement. By incorporating an elastic medium into a transducer of this type, a self-centring force is provided for the magnet after the load has been removed.

Tappin et al (1980) were the first to utilise this principle of transduction in a transducer for measuring local plantar shear stresses. Two non-magnetic stainless steel discs, 15.96 mm in diameter were used, which were bonded together with a layer of silicone rubber to produce a device 2.7 mm thick. The odd diameter was chosen to give the transducer a surface area of exactly 2 cm^2, which simplified the calculations made following calibration and data recording when voltage output was converted to stress. The magneto-resistor was glued onto a printed circuit board (pcb), which was recessed into one disc. This disc contained a groove machined perpendicular to the entry of the pcb. The magnet was centrally housed in the other disc, which had a ridge to match the groove of the disc containing the magneto-resistor. The interaction between the ridge and groove of the two discs confined the measurement of shear to one defined direction. In a prototype, the discs of this transducer were made from perspex because of its non-magnetic, non-conductive properties **(Pollard, 1984)**. However, this material was not strong enough during use and exhibited creep. Problems were also encountered with the power supply and signal output wires breaking at the points where they were soldered to the pcb just outside the transducer. To prevent this, the wires were first soldered to the pcb *inside* the transducer and then potted in an epoxy resin before being passed through small holes in the projecting pcb. The transducers were then linked by 10 m of cable to amplifiers and then to a ultra-violet (uv) recorder. Only a static calibration was performed before use. The transducer was placed in a circular recess and a cap was placed on top. A 5 kg weight was used to hold the transducer firmly and weights ranging from 0.5 kg to 4 kg were hung from a wire attached to the cap to cause shear. In a study of plantar shear stresses in different styles of shoe, **Pollard et al (1983)** taped six transducers to the foot beneath the hallux, first, second/third, fourth and fifth metatarsal heads, and the centre of the heel. Improvements to the electronic hardware for these transducers were made by **Tappin and Robertson (1991)**, who designed a portable unit which provided signal-conditioning and stored data for later downloading to a computer.

Laing et al (1992) produced a similar smaller transducer, 10 mm in diameter and only 1 mm thick. The wires entering this uni-axial transducer were made from flexible printed circuit to be more robust. A portable unit provided signal-conditioning, A/D conversion and data storage. Eight transducers could be interfaced, with the data logger having the capacity to store 8 seconds of data sampled at 128 Hz per transducer. Data was later downloaded from this portable unit to a computer. Again, these transducers were statically calibrated using a method similar to **Pollard (1984)**. At present there are no publications to describe the use of this transducer for in-shoe data collection. However, for preliminary barefoot measurements the transducers were stuck directly to the skin, beneath the hallux, the first, third and fifth metatarsal heads, and the medial aspect of the heel. In the case of measuring shear stresses, this method of mounting the transducers for in-shoe measurements has important consequences. Fixing shear transducers directly to the foot may result in the transducers remaining fixed in contact with the insock during forefoot loading as the underlying metatarsals slid forward and splayed. Thus any normal sliding of the skin against the insock would not influence the shear data. It is also likely that the sliding mechanism of the shear transducer would be hindered by being stuck to the foot.

A tri-axial transducer incorporating a bi-axial shear section was recently described by **Williams (1993)**. The shear section was assembled from two outer aluminium alloy discs, which each housed a magneto-resistor and had a groove; and a central acetal copolymer disc, which housed a magnet and on either side had a ridge (see figure 3.2). The discs were glued together with rubber between them to provide a self-centring force to the magnet. The interaction of the ridges and grooves of the discs confined the measurement of shear to two mutually perpendicular directions. The assembled device measured 15.96 mm in diameter and was 3.8 mm thick. An estimate of the resonant frequency of the transducer was evaluated by striking it parallel to one of its axes with a metal bar. Resonance occurred at approximately 1500 Hz, well above the relevant clinical range. Subsequent quasi-static calibration was carried out at a frequency of 1 Hz. For load cycles of ±5 N, ±10 N, ±20 N and ±50 N the accuracy of the transducer was found to be 9% of indicated stress. This transducer has been used for measuring stresses at the interface between the residual stump and prosthetic socket of amputee patients and was suggested for use in plantar stress studies (*op cit*).

miscellaneous techniques

A number of transducers for measuring in-shoe plantar stresses utilise principles of transduction that do not fall into one of the above groups. These transducers are described in this section.

The *Brand slipper sock* is a printing method which has been described by **Barrett (1976)**. The sock is made from polyurethane foam and impregnated with capsules of acid bromephenyl blue. Sodium bicarbonate is sprinkled on the sock before a test so that the acid bromephenyl blue develops in the alkaline medium when the capsules are ruptured during walking. Dark blue on the underside of the sock indicates areas of greatest pressure or shear, although it is difficult to distinguish between these two components of stress from the resultant print. This technique is not intended to provide quantitative results.

The Fuji Photo Film Company[16] have developed a pressure sensitive film similar in principle to the *slipper sock*. The Fuji *Prescale Film* is a two-part medium of an A and C-film. Microscopic bubbles are present on the A-film, which rupture on applied pressure. The C-film contains a medium which reacts with the colourless liquid in the A-film to produce a red stain. The films are available in four grades producing calibratable shades of red in the pressure range 0.5-130 MPa. An in-shoe study of ten patients has been conducted to assess this technique **(Ralphs et al, 1990)**. Unfortunately, the differences between the shades of red were found to be too subtle to distinguish pressure values accurately and in some cases the shades recorded were outside the calibrated range for the films. In addition, the films were loaded during

[16] Fuji Photo Film Company Ltd, 26-30 Nishiazabu, 2-chome, Minato-Ku, Tokyo 106, Japan.

walking, but were compared with supplied calibration films that had been derived after 2 minutes of static loading. These workers did attempt to load the films for about this time by conducting a test over thirty footsteps. Recently, an improved system for calibrating the films and for comparing test results to calibration films was described (**Liggins et al, 1992**). In a clinical setting this technique can be used quickly to obtain a composite map of maximum pressure over several steps. However, the high cost of the film is a limiting factor in its widespread use.

The most recently developed commercial system for plantar stress measurement is the *Parotec* by Kraemer[17]. Sixteen hydrocell sensors are mounted into an inlay at fixed locations. According to the manufacturers literature, the sensors can measure both pressure and shear in the range 2.5 kPa to 625 kPa (0.25-62.5 N/cm^2) to an accuracy of less than ±2.5%. Data is sampled at 100 Hz per sensor and is stored in a waist unit for later downloading to a computer. Early prototypes of this system incorporated over 100 sensors in an inlay. In reducing this large number of sensors to just 16 the manufacturers were satisfied that there was no significant data loss (personal communication). The sensor resolution does, however, seem inadequate for assessing patients with plantar pathologies that produce localised areas of high pressure.

A particularly novel design for a shear transducer was recently described by **Lebar et al (1992)**. The uni-axial device, 12.7 mm in diameter and 3.3 mm thick, consisted of two bronze discs, one with a wedge-shaped protrusion in the centre and the other with a light emitting diode (LED) and photodiode. The discs were held together with two vertical steel pins. The steel pins also provided a restoring force and by varying their stiffness the operating range of the device could be altered. When subjected to shear the wedge either increased or decreased the amount of light passing from the LED to the photodiode. The design of this transducer was later modified by this group to produce a device now 15 mm in diameter and 3.8 mm thick; and the steel pins were replaced by steel plates (**Lebar et al, 1993**). Only a static calibration was performed: linearity and hysteresis were 6.6% and 11.1%, respectively, for a maximum applied force of 22.3 N. The transducer was subsequently placed into an inlay and

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[17] Kraemer, Paromed Medizintechnik GmbH, D-8201 Markt Neubeuern, Germany.

measurements made beneath the hallux, first and fifth metatarsal heads, and the posterior aspect of the heel. Data was sampled at 200 Hz.

2.5.6 barefoot and in-shoe plantar pressures

generation of barefoot and in-shoe plantar pressures

The biomechanical processes whereby barefoot plantar stresses are generated and the modifications due to shoe factors, are described below with respect to observations from asymptomatic subjects.

Plantar pressures are generated by the bony anatomy of the foot acting through the plantar soft tissues when the foot is in contact with a supporting surface. The metatarsal heads, base of the fifth metatarsal and calcaneus are focal points for areas of localised high pressure.

The pattern of foot loading, as determined from the pressure recordings, reflects the general movements of the joints in the proximal and distal parts of the foot during the stance phase. In both barefoot and shod walking the foot is typically inverted towards the end of the swing phase, which results in the postero-lateral border of the heel making initial foot- or shoe-to-ground contact, respectively, at the beginning of the stance phase. Stability is provided as the heel is rapidly loaded and the heel pad, which is sandwiched between the calcaneus and supporting surface, compresses and flattens to distribute pressure over an area larger than the inferior aspect of the calcaneus.

At heel-contact the leg internally rotates and the foot plantarflexes and everts. Pressure is gradually registered in an anterior direction along the lateral border beneath the shaft of the fifth and possibly the fourth metatarsals. Quite often in pressure recordings a localised peak is noticeable beneath the bulbous base of the fifth metatarsal (**Hughes et al, 1991a**). Loading takes place gradually across the forefoot, from lateral to medial. In shoes with medial arch support the plantar surface beneath this area will also bear load (**Lord and Hosein, 1994**).

At heel-lift the body moves forward over the supporting leg and more load is transferred onto the forefoot. Discrete areas of high pressure may become visible beneath the ball area associated with the metatarsal heads. In barefoot walking, discrete areas of high pressure are typically visible beneath all five metatarsal heads. However, in a shoe with a narrow outsole and/or a tight fitting upper, it may be difficult to distinguish areas of high pressure beneath the fifth and/or the second and third metatarsal heads, respectively. In cases where the shoe insole provides significant cushioning, discrete areas of high pressure associated with a single metatarsal head may not be seen at all due to spatial smoothing.

As more load is transferred to the toes they dorsiflex at the metatarsophalangeal joints, which places the plantar aponeurosis and plantar skin beneath the ball under tension (**Bojsen-Moller and Lamoreux, 1979**). In barefoot walking the ball will be lifting off the ground at this stage and will be unloaded. In shod walking, however, dorsiflexion of the toes at the metatarsophalangeal joints is accompanied by the shoe flexing along the tread line, which will maintain a contact force between the ball of the foot and the shoe insole.

Push-off is typically from the hallux in barefoot walking. The sturdier form of the first metatarsal makes it better adapted to withstand axial compression as the foot inclines more to the horizontal (**Stokes et al, 1979**); and the hallux, supported by the tension in the flexor hallucis longus muscle, is able to provide greater leverage at push-off than the lesser four toes (**Bojsen-Moller and Lamoreux, 1979**). Maximum peak pressure beneath the hallux is typically registered beneath the antero-medial aspect, which reflects two simultaneous functions being carried out by the foot at push-off, ie. to provide the necessary leverage to propel the body forward and to transfer load medially to the contralateral foot. In shoes with a rigid outsole, dorsiflexion of the toes at the metatarsophalangeal joints will be restricted at push-off (*op cit*). It is envisaged that this will reduce the leverage provided by the hallux and will result in push-off from the medial aspect of the ball.

clinical findings from barefoot plantar pressure studies

Dynamic pressure measurements beneath the entire plantar surface allows foot function to be observed during gait. Changes in foot length and width during the stance phase; shifts in loading during gait revealing aspects of foot function; and the location of pressure peaks that may indicate an impending skin lesion, can be determined through this study method.

Unfortunately, from examining the publications of barefoot plantar pressure studies performed, it is apparent that no common approach has been employed by all researchers - likewise for in-shoe studies. The variety of measuring systems, calibration methods and analyses performed on data makes the comparison of results from different studies difficult. Even so, the values of pressure quoted and the results of other analyses that are presented are still important to note as long as the specific study methods employed and the study group characteristics are taken into account.

locus of the centre of pressure/force

The point of application of the instantaneous plantar pressure or ground reaction force during the stance phase is referred to as the centre of pressure (CoP) or the centre of force (CoF), respectively - both are the same. For a normal subject, the locus of the CoP shows a characteristic line-form that matches the parts of the foot that are in contact with the ground from heel-contact through midstance to push-off (figure 2.18).

Starting beneath the postero-lateral area of the heel at heel-contact, by 5% of the gait cycle the instantaneous CoP is beneath the centre of the heel. Between approximately 12% and 26% of the gait cycle the CoP progresses forward beneath the midfoot area slightly lateral to the midline. As the heel and then midfoot lift off the ground, increasing load is taken by the forefoot. Between 26% and 55% of the gait cycle the CoP traverses from lateral to medial beneath the ball area while still moving forward. At 57% of the gait cycle the CoP terminates beneath the hallux at push-off **(Elftman, 1939; Manley and Solomon, 1979; Hutton and Dhanendran, 1979; Katoh et al, 1983; Hutton and Stokes, 1991)** (figure 2.18). Slight variations on this general pattern have been observed in parts, possibly because of the influence of body sway (**Elftman; Grundy et al, 1975; Hutton and Dhanendran**). In some subjects, initiation of the CoP may occur beneath the proximal aspect or central part of the heel; progress forward slightly medial to the midline of the foot; travel beneath the second metatarsal head; and terminate beneath the interspace of the first and second toes.

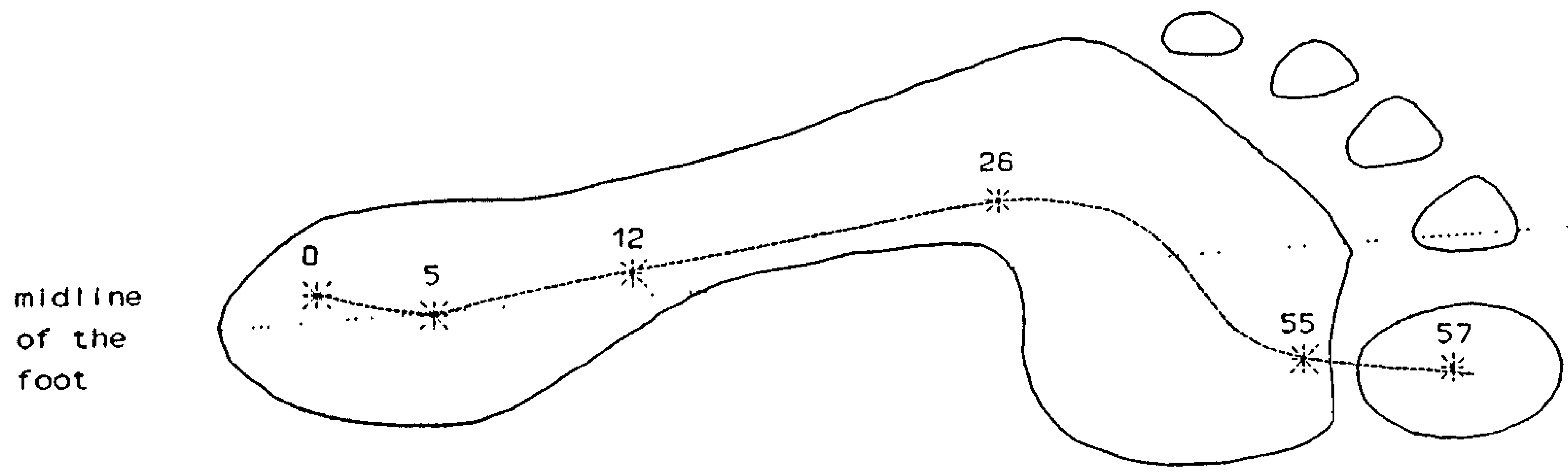

Figure 2.18. Typical path of the locus of the centre of pressure/force during barefoot walking. Time markers are indicated as a percentage of the gait cycle.

The spacing of time markers on a plot of the CoP gives an indication of the duration of weight-bearing. Between heel-contact and heel-lift, the CoP progresses smoothly and rapidly beneath the foot from the heel through the midfoot region. The clustering of time markers beneath the ball area indicates a greater period of weight-bearing here as more load is taken by the forefoot prior to the CoP moving rapidly from the ball area to the toes for push-off to occur (**Hutton and Dhanendran, 1979**).

foot-to-ground contact times

Floor mounted devices and discrete transducers have been used to determine the ground contact times of various parts of the foot during walking (**Scranton and McMaster, 1976; Hutton and Dhanendran, 1981; Soames, 1985**). Expressed as a percentage *of* (and not *into*) the stance phase, heel contact was found to last for between 45% and 65%. With heel contact maintained, weight shifts along the lateral border of the midfoot and into the forefoot. Midfoot contact lasts for between 49% and 58% and forefoot contact for between 63% and 87% of the stance phase. Weight transfer to the toes results in the hallux and the lesser four toes bearing weight for a similar length of time, 56% to 60% and 58%, respectively. The figures quoted here are from studies of mixed male and female groups, but in a comparison between the sexes,

Soames found no significant difference in the contact times for sixteen discrete areas analysed, for subjects wearing their own shoes. The styles of shoe worn by the subjects in this latter study were not, however, described, although they may have had a bearing on this result.

local plantar pressures

Recent studies of localised plantar pressure show wide variations, which are partly due to the variety of measuring devices that were used (table 2.3). In particular, discrete transducers had different surface areas; and pressures were averaged over large areas when measurements were made with the pedobarograph or a pressure-mat device.

The two largest plantar pressure studies to date both involved adults and children (**Hughes et al, 1990; Hennig and Rosenbaum, 1991**). Although the former study was larger and measured pressures at more plantar sites, toe function was of primary interest; in the latter study, a more comprehensive analysis of data was performed. **Hennig and Rosenbaum** studied one-hundred-and-eleven adults (mean age approximately 27 years) and fifteen infants (mean age approximately 2 years) and measured peak pressures at seven plantar sites. Comparing the two groups, adults:infants, the ratio of the mean of the peak pressures in these seven areas was found to be 3:1. One exception to this trend was observed in the midfoot where the difference in peak pressures between the groups was small. The magnitudes of peak pressure were also at their lowest beneath the midfoot, indicating the minimal weight-bearing role of this area. Reasons for the lower peak pressures in the infant group were attributed to the softer structure of the infant foot enabling forces to be distributed over a greater area; and the one-and-a-half times higher *body mass-to-foot contact area* ratio in the adult group. No correlation was found between body mass and peak pressure values, which was accounted for by a highly significant correlation between body mass and the load-bearing area during walking. Other workers have also found body mass and peak plantar pressures to be unrelated (**Cavanagh et al, 1991b; van der Zande et al, 1992**). Development of the longitudinal arch with ageing results in less and less load being taken by the lateral side of the midfoot and a bias towards more medial loading of the forefoot during gait as indicated by impulse analysis, ie. calculation of the local pressure-time integral (**Hennig and Rosenbaum**). For asymptomatic adults (over the age of 18 years), the different studies which are summarised in table 2.3 do not show a clear relationship between the magnitudes of local peak pressures and age. This is mainly due to the averaging of pressures over different areas, but even by intuition there would seem to be no correlation between these two quantities. Two studies have compared barefoot plantar pressures between the sexes. With the exception of the fourth toe, where significantly lower maximum peak pressures were measured in men compared to women (**Soames et al, 1985**), no significant difference in the local maximum peak pressures was found between the sexes to indicate any gender differences (*op cit*, **Holmes et al 1991**).

Table 2.3. Average maximum peak pressures beneath the asymptomatic foot during barefoot walking.

Source	Average maximum peak pressures (kPa)			
	hindfoot	midfoot	forefoot (mth)	toes
Schwartz et al (1964) 201 male subjects	1738 medial 949 lateral	-	591/1755/591 1st/3rd/5th	553 hallux
302 female subjects transducer area 1.27 cm^2	1050 medial 1032 lateral	-	921/1425/574 1st/3rd/5th	403 hallux
Betts et al (1980c) 50 feet, males and females, age range 17-56 years (estimation from graphs)	370	-	350/420/380/280/130 1st/2nd/3rd/4th/5th	440 hallux
Clarke & Cavanagh (1981) 27 subjects, transducer area 1 cm^2†	433 medial 391 lateral	59 medial 79 lateral	319 medial 324 lateral	379 medial 160 lateral
Grieve and Rashdi (1984)	208	88 anterior 47 posterior	163/212/197/160/97 1st/2nd/3rd/4th/5th	178 hallux
Pollard (1984) 10 male subjects, mean age 31.4 years, transducer area 1.76 cm^2	195	-	199/221 /160/125 1st/2nd-3rd/4th/5th	103 hallux
Soames (1985) 21 male subjects, mean age 23 ± 4.1 years	500 medial 403 lateral 781 posterior	148 anterior 94 posterior	538/531/594/461/398 1st/2nd/3rd/4th/5th	394/281/203/125/122 1st/2nd/3rd/4th/5th
11 female subjects, mean age 20.6 ± 3.3 years, transducer area 1.69 cm^2 (estimation from graphs)	466 medial 406 lateral 859 posterior	150 anterior 103 posterior	466/478/556/481/336 1st/2nd/3rd/4th/5th	475/313/250/203/122 1st/2nd/3rd/4th/5th

† These workers used the EMED System[9]. Only the dimensions of each loadcell were given and not the area over which pressures were averaged.

Table 2.3. (continued) Average maximum peak pressures beneath the asymptomatic foot during barefoot walking.

Source	Average maximum peak pressures (kPa)			
	hindfoot	midfoot	forefoot (mth)	toes
Zhu et al (1989) 5 male subjects, mean of data from the right and left foot (estimation from graphs)	390 anterior 400 posterior	-	360/440/350/279 1st/2nd/4th/5th	310 hallux
Hughes et al (1990) 160 subjects, age range 5-78 years (mean of data from the right and left foot)	183	-	213/258/242/192/133 1st/2nd/3rd/4th/5th	260/205/178/137/80 1st/2nd/3rd/4th/5th
Cavanagh et al (in press, cited in **Cavanagh and Ulbrecht, 1991)** 27 male subjects, middle-aged and elderly	345 medial 336 lateral	15 medial 113 lateral	319/533/ 446 1st/2nd/3rd-4th-5th	511/238/ 206 1st/2nd/3rd-4th-5th
Hennig & Rosenbaum (1991) 111 subjects, mean age 27 ± 8 years	312 medial 277 lateral	59	314/380/216 1st/3rd/5th	416 hallux
15 subjects, mean age 23.5 ± 5.7 months	119 medial 99 lateral	41	95 /99 /87 1st/3rd/5th	141 hallux
Holmes et al (1991) 10 male subjects	284	-	245/353/294/235 1st/2nd/3rd-4th/5th	373 hallux
10 female subjects, mean age of males *and* females 30 years (estimation from graph)	255	-	255/412/275/177 1st/2nd/3rd-4th/5th	412 hallux
Snow et al (1992) 45 female subjects, age range 22-55 years	-	-	236/345/269/186 1st/2nd/3rd/4th-5th	362 hallux
van der Zande et al (1992) 60 subjects (estimation from graphs)	429 medial 414 lateral	171	343 medial 450 central 243 lateral	343 hallux 171 2nd to 5th

The timing of pressure peaks at anatomically referenced sites has not been extensively published. For the one subject studied by **Betts et al (1980d)**, the forefoot was found to make ground contact as pressure peaked beneath the heel; and pressures beneath the metatarsal heads and hallux were found to peak at approximately the same time. **Hutton and Dhanendran (1981)** also observed pressures to peak at about the same time beneath the hallux and first metatarsal head, even though the hallux began bearing load some time after the first metatarsal head.

Only a few workers have measured pressures beneath all five metatarsal heads (see table 2.3). Many more have made measurements beneath selected metatarsal heads or have grouped two or more metatarsal heads together for analysing the peak pressure within that vicinity. Workers that have made individual measurements beneath the five metatarsal heads have found the second (**Betts et al, 1980d; Grieve et al, 1984; Hughes et al, 1990**) or third to be the site of highest maximum peak pressure (**Soames, 1985; Franks et al, 1983**).

The effect of walking speed on plantar pressures has notably been studied by two groups (**Shorten et al, 1989, Rosenbaum et al, 1994**). In both of these studies local peak pressures were generally found to increase with increasing walking speeds. However, exceptions to this trend were found beneath the second/third and fourth/fifth metatarsal head areas in the former study and beneath the fifth metatarsal head in the latter, suggesting push-off to occur from the medial side of the foot with increasing walking speed. Such a shift in loading is consistent with the work of **Bojsen-Moller and Lamoreux (1979)** who hypothesised push-off to occur about a transverse metatarsophalangeal joint axis (passing through the first and second metatarsal heads) during fast walking.

clinical findings from in-shoe plantar pressure studies

In contrast to numerous barefoot plantar pressure studies, there have been relatively few in-shoe plantar pressure studies. This section summarises the important findings of these in-shoe studies.

It is important that in-shoe plantar pressure studies take into account the characteristics of the shoes being worn. Shoe style, the materials used in construction and the fit each have individual effects on altering foot function. The greatest reductions in normal ranges of motion at the joints in the forefoot may occur as a result of a narrow shoe cramping the toes and a rigid sole impeding flexion at the metatarsophalangeal joints. Reductions in ranges of motion at the ankle and subtalar joints in the hindfoot may occur as a result of increased heel height.

The locus of the CoP in-shoe has been observed by one worker, but the path - an average calculated from the right foot of 11 subjects - was not plotted against an outline of the foot shape (**Rose et al, 1992**). Consequently, it appears that the locus of the CoP originates beneath the postero-lateral region of the heel; progresses forward beneath the midfoot lateral to the midline; traverses lateral to medial beneath the ball area while still moving forward (from just posterior of the third metatarsal head to just anterior of the first metatarsal head); and finally terminates beneath the hallux - apparently no different from barefoot walking. Time markers were not plotted on the locus of the CoP, so the duration of weight-bearing on different parts of the foot cannot be determined.

Wearing shoes tends to cause a shift in forefoot loading from the central to the medial metatarsal heads (**Pollard, 1984; Soames, 1985**). The impulse, ie. the force-time integral, is reduced beneath the heel and central metatarsal heads and increased beneath the medial metatarsal heads when shoes are worn (*op cit*). This indicates a rapid transfer of load from the heel to the forefoot, but increased weight-bearing beneath the medial side of the forefoot prior to push-off. Expressed as a percentage of the stance phase, heel, forefoot and hallucial contact with the shoe insole were found to last for approximately 68%, 80% and 82%, respectively (*op cit*, estimated from graphs). In shod walking, these areas of the plantar surface remain in contact with the shoe insole longer than they remain in contact with the ground during barefoot walking. It appears that as the forepart of the shoe comes down on to the ground and as the heel lifts off the ground, the sole of the foot in these respective areas is in contact with the shoe insole, ie. before and after significant shoe outsole-to-ground contact, respectively.

The results from the largest in-shoe plantar pressure studies are summarised in table 2.4 - studies of fewer than five subjects are not included. A comparison between these studies is made difficult for two reasons. Firstly because of the variety of analyses that were performed on the data; and secondly, because the shoes that were worn by the subjects were not described. It is immediately obvious that the pressures recorded by **Rose et al (1992)** are lower than those recorded by other workers. Particularly noticeable are the low pressures recorded beneath the forefoot. The reason for this is related to the combined analysis performed on the medial three and the lateral two metatarsal heads, compared to separate measures made beneath the metatarsal heads by other workers. By averaging pressures over larger areas, lower values are achieved. Some workers have allowed subjects to wear their own, quite different shoes during tests (**Soames, 1985**; **Rose et al**). Different styles and constructions of shoes may affect gait and subsequently plantar stresses. At the extreme, the effect of shoe style on plantar stresses is seen in high-heeled shoes in which peak pressures and shear stresses beneath the forefoot are almost doubled in comparison to shoes with a low heel (**Pollard, 1984**; **Snow et al,1992**).

Table 2.4. Average maximum peak pressures beneath the asymptomatic foot during shod walking.

Source	Average maximum peak pressure (kPa)			
	hindfoot	midfoot	forefoot (mth)	toes
Pollard (1984) 10 male subjects, mean age 31.4 years, transducer area 1.76 cm^2	145 central	-	171/173/107/88 1st/2nd&3rd/4th/5th	105 hallux
Soames (1985) 21 male subjects, mean age 23 ± 4.1 years	425 medial 450 lateral 500 posterior	90 anterior 145 posterior	525/470/410/300/220 1st/2nd/3rd/4th/5th	435 hallux 400/260/180/195 2nd/3rd/4th/5th
11 female subjects, mean age 20.6 ± 3.3 years, transducer area 1.69 cm^2 (estimation from graphs)	400 medial 430 lateral 525 posterior	80 anterior 175 posterior	570/380/395/270/220 1st/2nd/3rd/4th/5th	385 hallux 355/325/275/255 2nd/3rd/4th/5th
Zhu et al (1991) 10 male subjects, mean age 29.6 ± 5.8 years, transducer area 0.95 cm^2 right foot	435 anterior 579 posterior	-	480/573/427/241 1st/2nd/4th/5th	607 hallux
left foot	657 anterior 449 posterior	-	459/616/497/383 1st/2nd/4th/5th	419 hallux
Rose et al (1992) 5 subjects	98 medial 101 lateral	-	96 / 93 1st-2nd/3rd-4th-5th	74 hallux 47 second

It is considered important to note the findings of the one study which could be found in the literature comparing in-shoe plantar pressures between the sexes. With the exception of the third and fourth toes, where significantly lower maximum peak pressures were measured in men compared to women, **Soames (1985)** found no significant difference in the local maximum peak pressures between the sexes, indicating no gender differences.

2.5.7 barefoot and in-shoe plantar shear stresses

The biomechanical processes whereby barefoot and in-shoe plantar shear stresses are generated are described in this section along with the findings of barefoot and in-shoe plantar shear stress studies.

It is apparent from a review of the literature that only two studies provide comprehensive analysis of plantar shear stress data (**Pollard, 1984; Tappin and Robertson, 1991**). However, because of the mobile structure of the foot and the plane in which shear stress measurements are made, there is great potential for shear data to augment the present understanding of both barefoot and in-shoe foot function.

shear force

Because plantar shear stress measurement is a relatively new and unknown field, it is useful to precede a discussion of local plantar shear stresses with a general discussion of ground reaction forces and more specifically shear forces exerted by the body through the foot during walking.

A *forceplate* is used to measure ground reaction forces. Most modern types are electronic and consist of a flat, rigid plate with corner supports instrumented to typically measure three mutually perpendicular components of force exerted by the body. A force applied normal to the surface of the forceplate is defined as the *vertical* component. A shear applied to the forceplate in the direction of progression, the *longitudinal* direction, is defined as *anterior* shear and in the opposite direction as *posterior*. Shear on the forceplate in the direction perpendicular to the line of progression, the *transverse* direction, is defined as *medial* shear when directed towards the midline of the body, and as *lateral* shear when directed away from the midline of the body.

For a normal subject, the vertical component of the ground reaction force typically has a double-hump form (figure 2.19a). The first peak occurs when the heel has made full ground contact and the body is decelerating and the second peak occurs at push-off when the body is being accelerated forward. In the longitudinal direction there is an anterior deceleration force (relative to the forceplate), which has a peak magnitude of approximately 26% of bodyweight. At midstance the force in the longitudinal direction is reversed; and at push-off there is an acceleration force with a peak magnitude of approximately 25% of bodyweight (figure 2.19b). In the transverse direction there is a transient lateral force at heel-contact, which has a peak magnitude of approximately 9% of bodyweight (figure 2.19c). This is followed by a medial force which has a double-hump appearance and lasts until shortly before push-off: the first peak has a magnitude of approximately 14% of bodyweight and the second, approximately 9%. At push-off there is a transient lateral force, which has a maximum magnitude of approximately 4% of bodyweight **(Hamill et al, 1989)**.

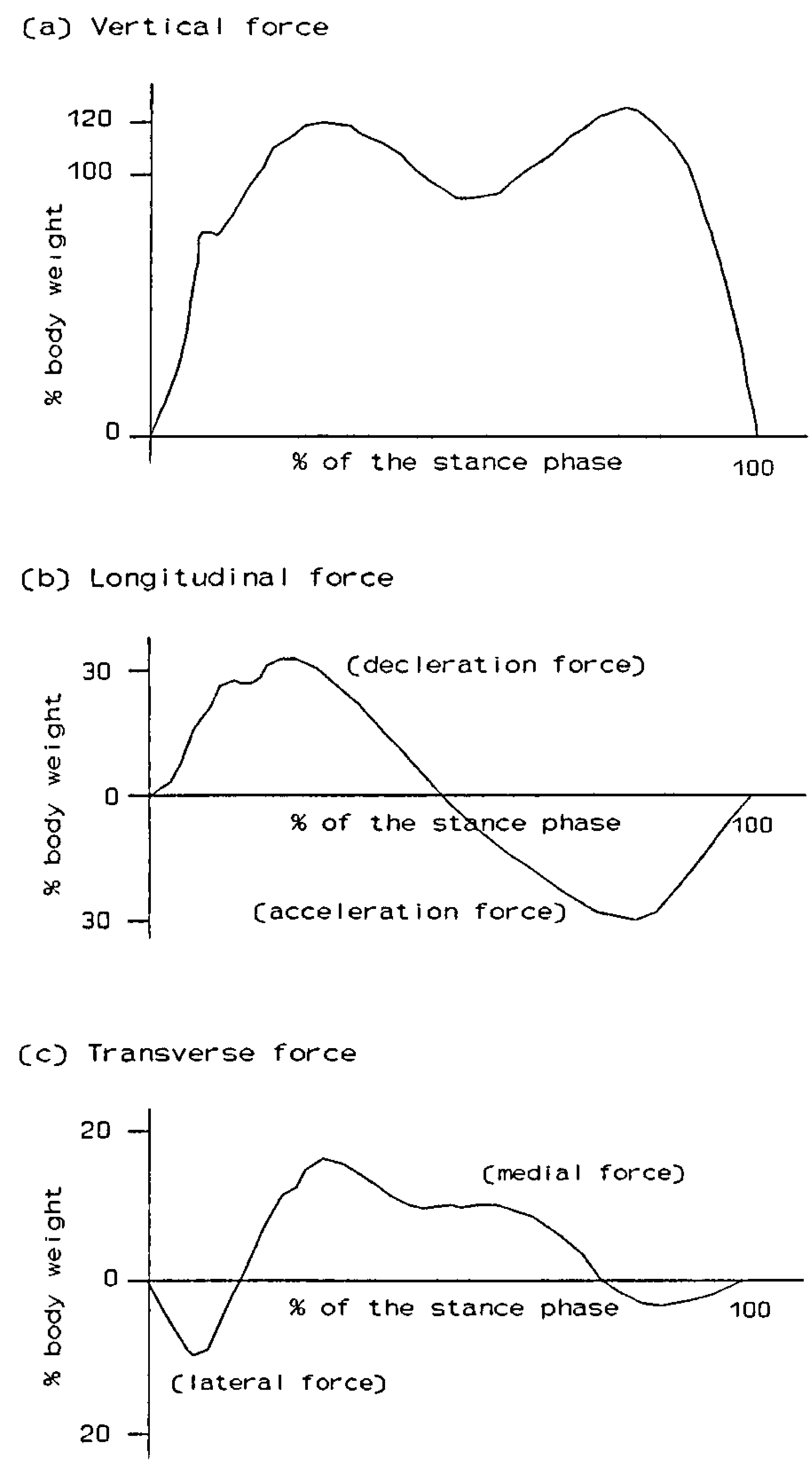

Figure 2.19. Typical forceplate output: (a) vertical (b) longitudinal and (c) transverse components of the ground reaction force (from ***Hamill et al, 1989****, with permission).*

generation and patterns of barefoot and in-shoe plantar shear stresses

Whereas the shear forces measured by a forceplate are exerted by the body through the foot to the ground, local plantar shear is generated by, and related to, the anatomy of the foot. The same terms used to describe the directions of shear on a forceplate are used to describe the directions of local plantar shear. For the forceplate these directional terms are described with respect to the direction of progression, whereas for local plantar shear they are described with respect to the midline of the foot. The longitudinal direction of local plantar shear is parallel to the midline. Most people tend to "toe-out" during walking so the longitudinal direction of local plantar shear is usually not parallel to the direction of progression.

With a knowledge of foot function and the typical shear output from a forceplate (figure 2.19b and c), it is possible to describe how barefoot and in-shoe plantar shear stresses may be generated. Articulations of the underlying bony structure of the foot; sliding of the foot relative to its supporting surface; deformation of the plantar soft tissues; and deceleration and acceleration of the body during walking are thought to be responsible for the generation of local plantar shear. However, the extent to which any one of these factors is responsible for the generation of barefoot and in-shoe plantar shear is not known. If it is assumed that the foot does not slide relative to its supporting surface during walking, as a result of a suitable frictional contact, the generation of local plantar shear will be largely dependent on: (1) bony articulations, particularly those that result in flattening and splaying of the metatarsals; and (2) deceleration of the body between heel-contact and foot-flat and acceleration of the body at push-off. With respect to the bony anatomy of the foot and with reference to the typical shear output from a forceplate (figure 2.19), the following movements of the foot and bony articulations are thought to occur and consequent patterns of local plantar shear are thought to be generated with respect to the supporting surface.

The foot is typically inverted at heel-contact. At heel-contact a postero-lateral directed shear is thought to be generated beneath the heel followed rapidly by an antero-medial directed shear of greater magnitude as the body is decelerated (see figure 7.1 as an aid to interpreting the resultant directions of shear). Once the forefoot or forepart of the shoe outsole has made contact with the ground, the foot will evert at the subtalar joint which will increase the range of motion at the midtarsal joint and cause flattening and splaying of the metatarsals about the second toe (**Shereff et al, 1990**). Weight-bearing extension of the foot at midstance is thought to be more pronounced beneath the medial rather than the lateral longitudinal arch - the lateral longitudinal arch is already very low to the ground and has little capacity to lower further. An antero-lateral directed shear is thought to be generated beneath the third, fourth and fifth metatarsal heads: the peak magnitude of the anterior component is expected to be greater beneath the third followed by the fourth and fifth metatarsal heads; while the peak magnitude of the lateral component is expected to be greater beneath the fifth followed by the fourth and third metatarsal heads. Only an anterior component of shear is thought to be generated beneath the second metatarsal head as this metatarsal flattens in the sagittal plane (figure 2.20). An antero-medial directed shear is thought to be generated beneath the first metatarsal head: the peak magnitude of the anterior component is expected to be less than the peak anterior shear generated beneath the second metatarsal head, but greater than the peak magnitude of the anterior component of shear generated beneath the third, fourth and fifth metatarsal heads. Also during midstance, loading of the foot is thought to cause the calcaneus to slide slightly backwards to generate a posterior directed shear beneath the heel (figure 2.20).

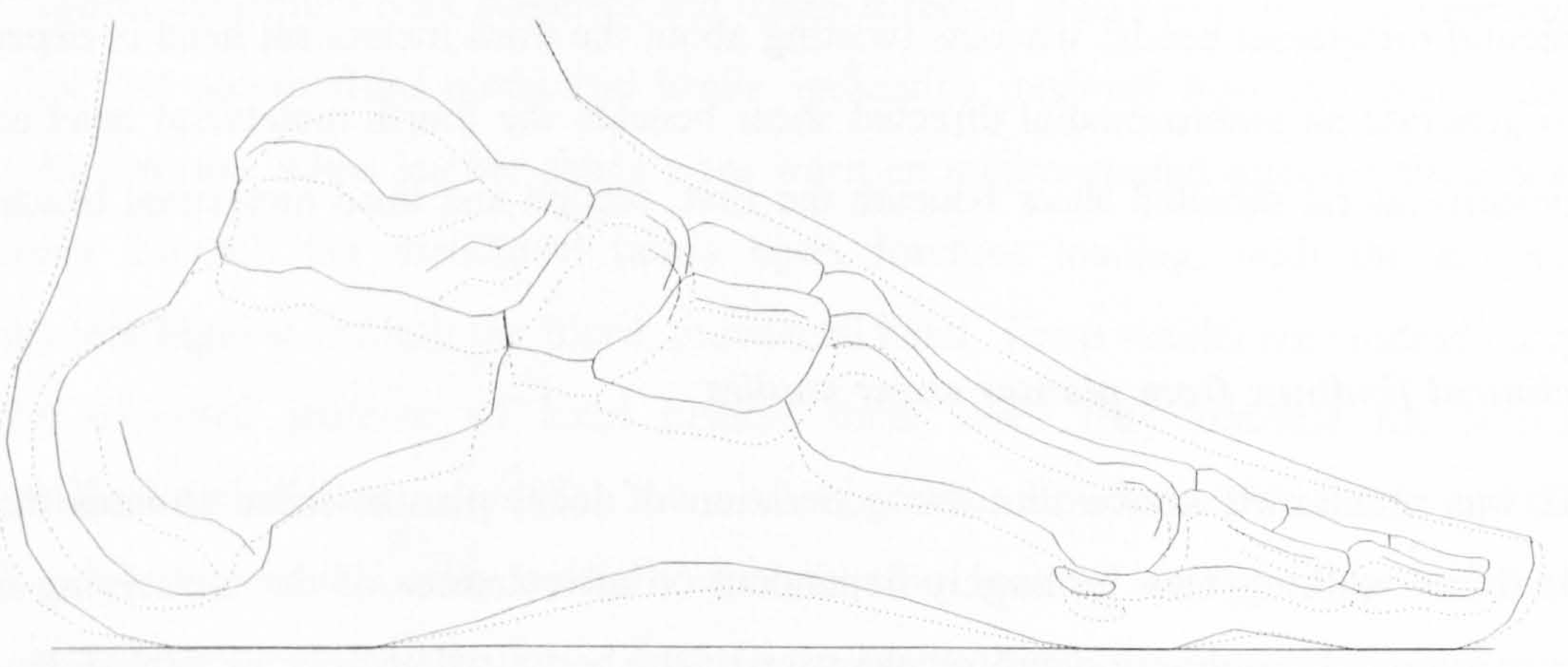

Figure 2.20. Change in foot shape upon weight-bearing: broken line indicates the loaded shape of the foot during midstance.

At heel-lift the toes will extend at the metatarsophalangeal joints and the shoe will flex along the tread line. A forward shift in loading will increase the weight-bearing role of the toes, which will hold them in opposition to the supporting surface. The direction of shear generated beneath the first metatarsal head is thought to reverse at this stage as the proximal end of the first metatarsal swings upward and the head rotates in situ on the sesamoids, being prevented from slipping forward by the rigid hallux (**Price, 1959**, see figure 7.8a). In contrast, as the proximal ends of the lesser metatarsals swing upward their heads will rotate and translate forward (*op cit*, see figure 7.8b). Buckling of the toes, ie. extension at the metatarsophalangeal joints and flexion at the proximal interphalangeal joints, allows this translation. Shear generated beneath the lesser metatarsal heads is thought to reverse direction once their forward translation has stopped and as the proximal ends of the metatarsals continue to swing upward.

Forceplate data indicates a posterior directed shear and a torque are generated at push-off (**Schoenhaus et al, 1979**). The torque indicates internal rotation of the leg with subsequent twisting about the foot, which is thought to occur at the site of highest maximum peak pressure beneath the ball - either the second or third metatarsal head. Twisting about the second metatarsal head as the proximal ends of the metatarsals swing upward is expected to generate an antero-medial directed shear beneath the third and fourth metatarsal heads and a postero-lateral directed shear beneath the first and second metatarsal heads; whereas twisting about the third metatarsal head is expected to generate an antero-medial directed shear beneath the fourth metatarsal head and a postero-lateral directed shear beneath the first, second and third metatarsal heads.

clinical findings from plantar shear studies

It was mentioned above that the generation of local plantar shear stresses during barefoot walking may be largely dependent on articulations of the underlying bony structure and deceleration and acceleration of the body, rather than the foot sliding. In several studies of barefoot plantar shear stresses workers have taped transducers directly to the sole of the foot (**Pollard, 1984; Tappin and Robertson, 1991; Laing et al, 1992**). Sliding is likely to occur at heel-contact and during forefoot loading when transducers are used in this way, especially if the transducers have a smooth under surface and slightly elevate the parts of the foot where they are placed. In contrast, the foot is less likely to slide inside a correctly fitting shoe when this methodology is used and flattening of the arches and splaying of the metatarsals with forefoot loading may be limited also. With these ideas in mind, the shear stresses generated beneath the forefoot could reasonably be expected to be higher during barefoot walking. It is surprising to discover then that experiment has not confirmed this.

Shear stresses were measured beneath the heel, hallux and the first, second/third, fourth and fifth metatarsal heads by **Pollard (1984)** and a sample set of shear records were presented for one subject. During barefoot walking, a high frequency postero-lateral directed shear (relative to the transducer) was recorded beneath the heel at heel-contact, followed rapidly by a larger antero-medial directed deceleration shear. As loading progressed to the forefoot, a postero-lateral directed shear was recorded beneath the first and second/third metatarsal heads and an antero-lateral directed shear was recorded beneath the fourth and fifth metatarsal heads. This is an indication of twisting beneath the forefoot, but is unexpected between foot-flat and midstance when eversion of the foot on loading is known to cause flattening of the arches and splaying of the metatarsals about the second metatarsal (**Shereff et al, 1990**). Loading of the toes resulted in an antero-lateral directed shear beneath the hallux. At push-off, a postero-lateral directed shear was recorded beneath the hallux and metatarsal heads. The highest maximum peak posterior and lateral directed shears were recorded beneath the first and second/third metatarsal heads, indicating push-off from the medial ball area. In contrast, when leather shoes were worn an antero-medial directed shear was recorded beneath the metatarsal heads upon forefoot loading, with the anterior component highest beneath the fourth metatarsal head. These results are contradictory to the expected patterns of local plantar shear since they indicate the lateral longitudinal arch flattens more than the medial between foot-flat and heel-lift; and the medial four metatarsals splay medially upon forefoot loading instead of just the first. In both barefoot and shod walking, shear in the longitudinal direction was reversed as the heel lifted. The peak posterior and lateral directed shear stresses that were recorded beneath the metatarsal heads were higher than the peak shear stresses that were recorded in the anterior and medial directions for both barefoot and shod walking, with the exception of the fourth metatarsal head in-shoe (**Pollard**). In a barefoot study by **Tappin and Robertson (1991)**, plantar shear stresses were found to cease altogether beneath the hallux at 94% of the stance phase, which implies the final motion at push-off to be an elevation of the foot rather than a backward thrust. The values of average "peak-to-peak"[18] longitudinal and transverse shear stresses presented by **Pollard** for measurements made beneath the asymptomatic foot during barefoot and shod walking are summarised in table 2.5. For both directions of shear, the "peak-to-peak" magnitudes measured locally were lower in barefoot walking.

*Table 2.5. "Peak-to-peak" longitudinal and transverse shear stresses (kPa) beneath the asymptomatic foot during barefoot and shod walking, as measured by **Pollard (1984)** (adapted, mean of ten subjects).*

Site	Barefoot		Wearing leather shoes	
	longitudinal shear	transverse shear	longitudinal shear	transverse shear
hallux	59	29	43	27
mth1	87	56	57	36
mth2/3	175	69	65	55
mth4	95	74	62	46
mth5	66	30	50	27
heel	101	47	62	34

2.6 DIABETES AND DIABETIC FOOT COMPLICATIONS

2.6.1 introduction

Diabetes mellitus is a disease characterised by a raised blood glucose concentration due to a deficiency or diminished effectiveness of the essential hormone insulin. There are two types: *non-insulin dependent diabetes mellitus (niddm)*, for which patients regulate their blood glucose level with an appropriate diet or an appropriate diet plus tablets (oral hypoglycaemic agents); and *insulin dependent diabetes mellitus (iddm)*, for which patients regulate their blood glucose level with daily insulin injections and an appropriate diet.

[18] **Pollard (1984)** presented *peak-to-peak* values of the local longitudinal and transverse plantar shear stresses. In the terminology of this thesis *peak-to-peak* is equivalent to *maximum peak-to-maximum peak*.

The complications associated with longstanding diabetes can lead to a host of pathologies. Conditions such as nephropathy (renal failure), retinopathy (blindness), vascular disease and heart disease are more common in patients with diabetes than in non-diabetics. Foot problems though are the most significant complication of the disease (**Frykberg, 1991**). In the UK more hospital beds are occupied by patients with diabetes who have foot problems than all other complications of diabetes combined (**Malins, 1968**). The associated annual cost to the National Health Service for this inpatient care has been conservatively estimated to be £223.4 million (**Laing et al, 1991a**). Of those pathologies that affect the foot the plantar ulcer is perhaps the most prevalent (**Frykberg**). Treatments that reduce hospital stays by managing these lesions on an out-patient basis or prevent them occurring are therefore preferred.

Ulcers on the plantar surface are typically found in patients with peripheral neuropathy, a condition characterised by sensory loss, dry skin and small muscle paralysis. Sensory loss plays a vital role in predisposing to the development of these lesions (**Masson et al, 1989**). However, a patient "at risk" may live for many years without developing an ulcer due to the avoidance of specific precipitating factors, the most important of which is believed by clinicians to be mechanical trauma.

2.6.2 diabetic peripheral neuropathy

Peripheral neuropathy (or simply *neuropathy*) is a common nerve complication of diabetes mellitus that is apparently related to the intensity (degree of hyperglycaemia, ie. the level of glucose in the blood) and duration of the disease (**Pirart, 1978**). A strong correlation with age and a predominance among male patients has also been found (**Greene et al, 1988**). In diabetes all components of the lower extremity nerves may become affected: *sensory*, *motor* and *autonomic*.

Somatic sensorimotor neuropathy is the most common and recognizable type of nerve disorder in patients with diabetes (**Frykberg, 1991**). Both sensory and motor nerves are affected. Sensory involvement begins *distally* in the toes and progresses *proximally* (**Dyck, 1987**) leading to loss of pain and thermal sensation; impaired or absent knee and ankle reflexes; and reduced vibration perception and light touch sensation. The inability to perceive pain predisposes the foot to injury from burns, and mechanical injury eg. from treading on sharp objects or the repeated forces of gait. Denervation of the motor fibres to the intrinsic muscles of the foot causes weakness and wasting of the musculature. Paralysis of the extensor digitorum brevis, lumbrical and interosseous muscles cause the metatarsophalangeal joints to become hyperextended and the interphalangeal joints flexed. Subsequent claw-toe deformities lead to anterior displacement of the sub-metatarsal fat cushions and prominent metatarsal heads (**Bojsen-Moller, 1979b**). Atrophy of plantar adipose tissue caused by denervation of the fat pads of the heel and the sub-metatarsal fat cushions may also lead to bony prominences (**Gooding et al, 1986, Jahss et al, 1992b**).

The autonomic nervous system functions without conscious control to regulate heart rate, blood pressure, blood flow, body temperature and other physiological processes. Impaired sweating, resulting in dry skin; and marked increases in blood flow compared to patients without diabetes are characteristics of *autonomic neuropathy* (**Archer et al, 1984; Corbin et al, 1987; Irwin et al, 1988**). Pure sensorimotor or autonomic neuropathy is unusual (**LeQuesne et al, 1991**): a close association exists between the two (**Foster and Edmonds, 1987a**).

2.6.3 diabetic neuropathic plantar ulceration

The earliest description of a diabetic neuropathic plantar ulcer was made by **Mott** in 1818 (cited in **Croft et al, 1982**). These ulcers were recognized as a complication of diabetes before 1870 (**Faris, 1991**), but it was not until the end of the 19th Century that hypotheses concerning their aetiology were published. **Treves (1884)** believed diminished "nutrition[19] of certain parts" of the foot; "dulled" sensation of the sole; and the effects of "pressure upon the skin" were "the chief factors in the production of perforating ulcers". He thought "disturbance of nutrition" was the most important aetiological factor after observing plantar ulcers in patients without "perverted sensation in the limb" and "nothing peculiar in the pressure to which the part" was subjected. **Pryce (1887)** believed "locomotor ataxy" (motor neuropathy) and

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[19] "nutrition" in this context most likely refers to a reduced blood supply, with a subsequently reduced supply of oxygen and nutrients, such as glucose, to the tissues.

"peripheral neuritis" (sensory neuropathy) to be causative factors in ulceration; and noted that the previously ulcerated feet of patients with diabetes sometimes showed a "glossy skin" - a suggestion of autonomic neuropathy.

predisposing, precipitating and aggravating factors in ulceration

The factors that are now considered responsible for the development of plantar ulcers in the insensitive diabetic foot may be classified as *predisposing* and *precipitating.*

Complications of diabetes are grouped as predisposing factors. Peripheral neuropathy and connective tissue changes are the two most important of these to place the foot "at risk". Detrimental connective tissue changes are related to a patient's control of their diabetes. Glycohaemoglobin, a measure of poor diabetic control, has been found to be higher in patients with ulcers than those without (**Croft et al, 1982**). Nonenzymatic glycosylation of collagen (**Schnider and Kohn, 1980**) and keratin (**Delbridge et al, 1983**) has been related to poor diabetic control (**Yue et al, 1983**), which is significant in that glycosylation of collagen produces an increase in intermolecular crosslinking making the skin and subcutaneous tissues rigid and inflexible (**Hamlin et al, 1975**). It is thought glycosylation produces a similar structural change in keratin (**Delbridge et al, 1985**).

Despite the presence of predisposing factors, ulcers do not usually develop without the influence of precipitating factors, which act to cause actual injury. Repeated mechanical trauma from walking is a common threatening cause of ulceration. It has been hypothesised that rigidity of the skin and subcutaneous tissues and the build up of glycosylated keratin on the plantar surface in response to pressure, predisposes to ulceration through inadequate pressure distribution during walking (**Frykberg, 1991**). Foot deformity and limited joint mobility are also thought to accelerate damage to the foot from repeated mechanical trauma. For example, plantar ulceration may occur beneath a prominent metatarsal head. It has also been noted that patients with diabetes who have a forefoot varus deformity, ie. a fifth metatarsal head which is more plantar than the first, often ulcerate beneath the fifth metatarsal head (**Mueller et al, 1990; Schoenhaus et al, 1991**). Assessing structural foot pathologies such as this may help to predict an area at risk of ulcerating.

An association between abnormally high plantar pressures and limited mobility at the metatarsophalangeal and subtalar joints of the insensitive diabetic foot has been found by a number of workers (**Delbridge et al, 1987, Cavanagh et al, 1991a, Fernando et al (1991)**. In these patients, neuropathy is thought to be the critical factor in causing ulceration. This is backed by observing subjects who have limited joint mobility at the first metatarsophalangeal joint, ie. having *hallux rigidus*, who have a tendency to develop an area of callus on the medial aspect of the first metatarsophalangeal joint, but do not go on to ulcerate. It is also important to note that plantar ulceration may also be caused by other events, such as burning of the skin, for example from a hot water bottle; or by chemical attack, for example from corn and wart removal preparations containing salicylic acid.

Peripheral neuropathy is also classified as an *aggravating factor* in ulceration along with infection. These factors may not act to increase the amount of damage done to tissue and bone, but either way they delay healing.

development of the diabetic neuropathic plantar ulcer

The most frequent cause of ulceration brought about by mechanical factors is thought to be the neglected callosity (**Edmonds, 1986**). Callus forms in protective response to locally high pressures. In the clawed-toe deformity seen in patients with diabetes who also have sensorimotor neuropathy, the metatarsal heads are made prominent and callus may form on the plantar skin beneath them. With excessive callus build-up, the compliant subcutaneous tissue between the metatarsal heads and callus becomes subjected to high local stresses during weight-bearing (**Thompson, 1988**), but the pain associated with this goes unnoticed by the patient (figure 2.21a). Breakdown of the trapped subcutaneous tissue results in a cavity filled with plasma and blood (figure 2.21b). This cavity gradually enlarges and eventually breaks through the skin to form the ulcer (**Delbridge et al, 1985**) (figures 2.21c and 2.22). For some patients, amputation becomes necessary if bone is infected through the open site from micro-organisms present on the skin or within footwear.

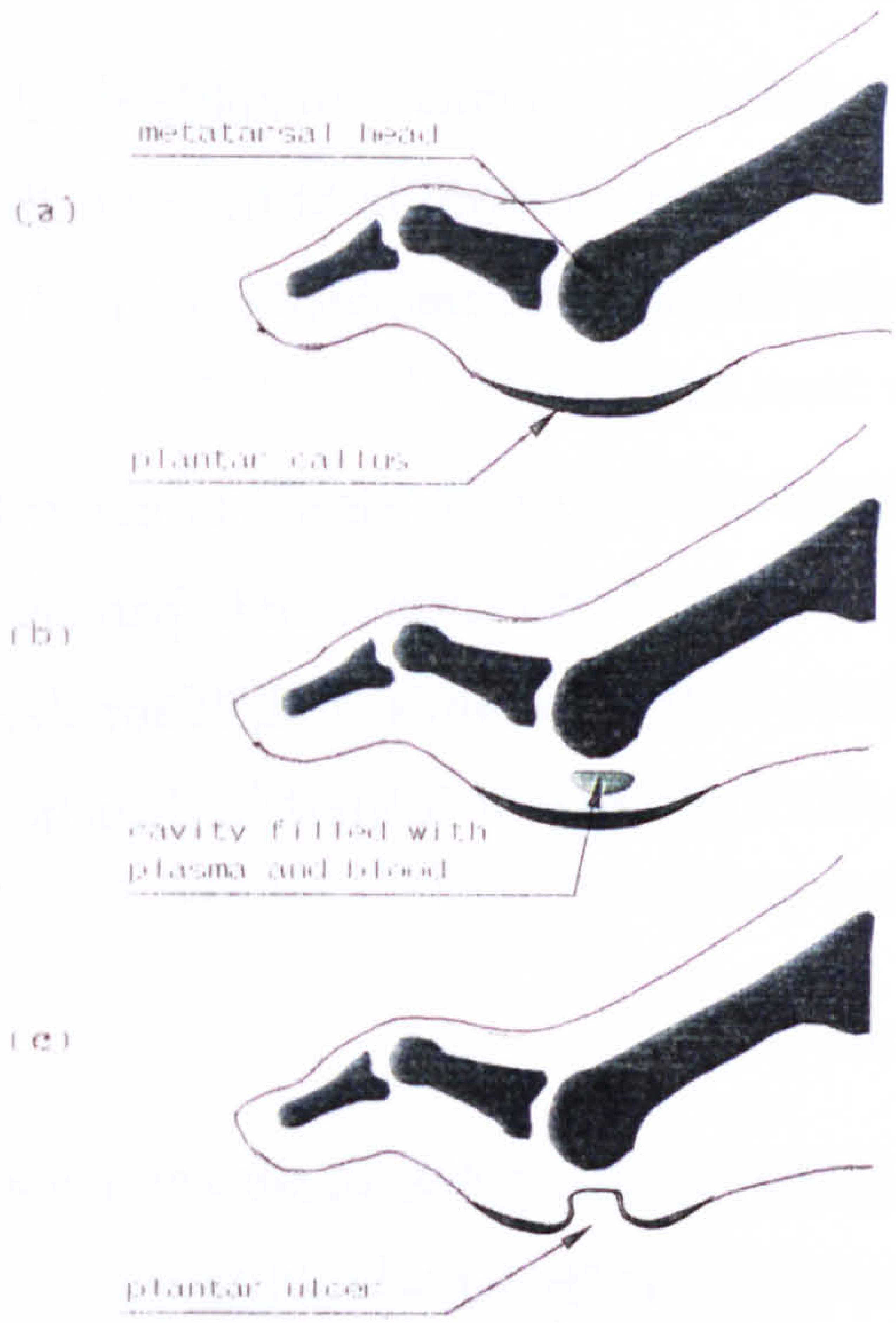

Figure 2.21. Stages in the development of a forefoot ulcer beneath the insensitive diabetic foot.

Figure 2.22. Typical appearance of a neuropathic plantar ulcer.

2.6.4 clinical findings from plantar stress studies

In dynamic plantar stress studies, patients with diabetes have been the most extensively studied patient group. A greater proportion of these studies have documented plantar pressure while regarding the shear component as clinically insignificant.

The apices of the toes and the plantar aspects of the first and third metatarsal heads are the most common sites of plantar ulceration in patients with diabetes (**Holstein et al, 1976; Ctercteko et al, 1981; Edmonds et al, 1986; Veves et al, 1992**). As a result, the forefoot has been the main area of interest in diabetic plantar stress studies.

barefoot measurements

The supposition that plantar ulcers develop at sites of abnormally high pressure has so far been supported by one prospective study. In the period of follow-up of eighty-six patients, **Veves et al (1992)** noted twenty-one plantar ulcers had developed, eight of which were at sites where abnormally high peak pressures (>1.23 MPa) had been earlier measured. Several retrospective studies have found maximum peak pressures to be coincident with the sites of previous ulceration; and significantly higher pressures beneath the forefoot of patients with neuropathy and healed plantar ulcers, compared to patients with neuropathy and no history of ulceration - a finding which is partly related to reduced toe loading due to the presence of claw-toe deformities (**Barrett and Mooney, 1973; Stokes et al, 1975; Ctercteko et al, 1981; Boulton et al, 1983; Duckworth et al, 1985; Cavanagh et al, 1987b; Smith et al, 1989**). With these findings in mind it should come as no surprise then that re-ulceration is a common occurrence in these patients.

Two groups have measured the *forces* beneath the feet of patients with diabetes who also have neuropathy. **Stokes et al (1975)** used a forceplate consisting of twelve beams to make measurements beneath the feet of patients with a history of plantar ulceration. For the four feet studied, a range of "highest maximum load" was given: from 175 N to 264 N (estimated from graph). Significantly lower toe loading and more lateral forefoot loading was also observed in this group compared to a group of asymptomatic subjects. **Ctercteko et al (1981)** used a higher resolution forceplate consisting of one-hundred-and-twenty-eight, 15 mm x 15 mm loadcells for measurements. This system allowed the forces at the sites of healed ulcers to be measured. Sites of healed ulcers were found to be coincident with the sites of maximum load. For the twenty-four patients with a history of plantar ulceration that were studied, the "mean peak force" was 228.7 N. Forces were also measured beneath the feet of patients with neuropathy and no history of ulceration and beneath the feet of asymptomatic subjects. In a comparison of all three groups, a considerable overlap in the ranges of the forces measured was found. Reduced toe loading in comparison to asymptomatic subjects was also observed in this study, but in contrast to the finding of **Stokes et al**, more medial forefoot loading was observed in patients that had previously ulcerated.

A number of workers who have measured local maximum peak pressures beneath the feet of patients with diabetes and asymptomatic subjects have sought to define a threshold of 'normality' (**Boulton et al, 1983; Smith et al, 1989**). The aim of this being to determine which patients may be at immediate risk of plantar ulceration. However, with the wide range of reported peaks measured beneath the sites of healed diabetic neuropathic ulcers (table 2.6), which in some cases overlap with the ranges reported for patients with diabetes and no foot problems and asymptomatic subjects, it is clearly difficult to do so.

Table 2.6. Maximum peak pressures measured beneath the forefoot sites of healed diabetic neuropathic ulcers during barefoot walking.

Source	Maximum peak pressures (kPa)
Boulton et al (1983)	> 1100
Boulton et al (1984)	920 to 3730
Smith et al (1989)	1060 ± 590
Pollard (1984)	200 to 500
Cavanagh et al (1987b)	312 to 1895

The low maximum peak pressures recorded at anatomical sites by **Proano et al (1992)** was the result of spatial averaging. For the toe, metatarsal, tarsal and heel areas, considerable overlaps were found in the maximum peak pressures measured beneath the previously ulcerated and non-ulcerated contralateral feet of ten patients; and the feet of the patients and ten asymptomatic subjects. However, in an analysis of *average* maximum peak pressures, a significant difference was found beneath the metatarsal area between the feet of the patients that had previously ulcerated and the feet of the asymptomatic subjects. Although it was not clearly stated it appears that the sites of healed ulcers were predominantly beneath the ball area of the foot, which accounts for this result.

It is interesting to note that one group of workers has made a comparison between two systems for measuring plantar pressures (**Young et al, 1992b**). In the study of 20 patients with diabetes, with measurements made using the *Musgrave Footprint* and the *Optical Pedobarograph* (OPG), no significant differences were found in the maximum peak pressures measured beneath the hallux and heel, but maximum peak pressures were significantly higher beneath the first and fifth metatarsal heads when measured by the OPG. One of the reasons for these disparities was wrongly attributed to the lower measurement range of the Musgrave system, but it was correctly noted that the lower pressures measured by the Musgrave system were due to the larger sensors which led to greater spatial averaging.

One worker has measured shear stresses beneath the diabetic foot, but *only* in the longitudinal direction (**Pollard, 1984**). This inevitably underestimated the *true* local maximum peak shear stresses, which are vector-additions, ie. the *resultant*, of longitudinal (antero-posterior) and transverse (medio-lateral) components. Longitudinal shear stresses were higher beneath the sites of healed ulcers compared to intact sites beneath the patient's foot and corresponding sites beneath asymptomatic feet. The values of "peak-to-peak" longitudinal shear stresses presented ranged from 65 kPa to 180 kPa (0.65 $kgcm^{-2}$ to 1.80 $kgcm^{-2}$) beneath seven healed ulcers compared to 57 kPa to 65 kPa (0.57 $kgcm^{-2}$ to 0.65 $kgcm^{-2}$) for the corresponding sites beneath the asymptomatic feet. Four out of the seven sites of previous plantar ulceration was the site of highest maximum peak pressure beneath the foot; and five out of the seven sites was the site of highest "peak-to-peak" longitudinal shear stress beneath the foot. However, because no patient graphs were presented, it is impossible to know whether the anterior or the posterior directed shear stresses was higher for the sites that were studied.

in-shoe measurements

Although patients with diabetes who also have neuropathy often first ulcerate while wearing conventional shoes, there have been very few studies documenting the dynamic stresses beneath the feet of these patients inside such shoewear.

Two groups have found local plantar pressures to be only slightly reduced when patients wore conventional leather shoes compared to walking barefoot. Local pressures therefore remain high and so conventional shoes offers little therapeutic aid. In a study of six patients (seven feet) by **Pollard (1984)**, average maximum peak pressures at six plantar sites (hallux, heel and the first, second/third, fourth and fifth metatarsal heads), excluding the sites of healed ulcers, ranged from 71 kPa to 327 kPa (0.71 $kgcm^{-2}$ to 3.27 $kgcm^{-2}$) when walking in leather shoes compared to 36 kPa to 336 kPa (0.36 $kgcm^{-2}$ to 3.36 $kgcm^{-2}$) when walking barefoot (table 2.7); while maximum

Table 2.7. Average maximum peak pressures and "peak-to-peak" longitudinal shear stresses beneath the feet of patients with diabetes who have a history of plantar ulceration (measurements not made beneath the sites of healed ulcers): barefeet, shod and using walking casts, as measured by ***Pollard (1984)*** *(adapted).*

Site	Pressure (kPa)			Longitudinal Shear Stress (kPa)		
	barefoot	leather shoes	walking cast	barefoot	leather shoes	walking cast
hallux	36	71	-	29	24	-
mth1	336	327	129	175	107	30
mth2/3	253	226	46	143	96	36
mth4	201	187	64	128	79	22
mth5	156	113	50	56	45	16
heel	155	137	92	78	62	22

peak pressures measured at the sites of healed ulcers ranged from 220 kPa to 500 kPa (2.2 $kgcm^{-2}$ to 5 $kgcm^{-2}$) shod and from 200 kPa to 500 kPa (2 $kgcm^{-2}$ to 5 $kgcm^{-2}$) barefoot. Longitudinal shear stresses were also recorded by this worker, which were found to be lower in patients walking in shoes compared to walking barefoot: values ranged from 24 kPa to 107 kPa (0.24 $kgcm^{-2}$ to 1.07 $kgcm^{-2}$) shod and 29 kPa to 175 kPa (0.29 $kgcm^{-2}$ to 1.75 $kgcm^{-2}$) barefoot (table 2.7). Walking in shoes as opposed to barefoot resulted in lower maximum peak pressures and lower "peak-to-peak" longitudinal shear stresses at the sites of five out of seven sites of previous plantar ulceration. Three out of five of these sites was the site of highest maximum peak pressure beneath the barefeet of the patients and remained so when shoes were worn; while four out of five of these sites was the site of highest "peak-to-peak" longitudinal shear stress beneath the barefeet of the patients and remained so when shoes were worn. **Smith et al (1989)** was rather less thorough in his analysis of data in a study of eleven patients. A group average of the maximum peak pressure measured at seven plantar sites beneath each foot was simply quoted. This value was 610 ± 180 kPa (6.1 ± 1.8 $kgcm^{-2}$) for measurements made in-shoe and 670 ± 220 kPa (6.7 ± 2.2 $kgcm^{-2}$) for barefoot measurements. From the standard deviation of the means it is apparent that there was a considerable overlap between the in-shoe and barefoot plantar pressures measured.

2.6.5 management of the diabetic neuropathic foot

Management of the diabetic neuropathic foot has three purposes: to prevent primary ulceration; to aid healing in patients who have ulcerated; and to prevent re-ulceration. All three of these processes may be aided by chiropody treatment and the use of special shoes that are often bespoke.

chiropody and walking cast treatment of diabetic neuropathic ulcers

In the foot that *has not* ulcerated, callus is debrided to reduce locally high pressures and minimise the risk of ulceration **(Young et al, 1992a)**. There are a number of ways in which the chiropodist can treat the diabetic foot that *has* ulcerated. In cases of primary ulceration due to continued weight-bearing, the initial treatment is in three stages: firstly, callus is removed from around the open site, as far back as soft pink skin; secondly, the area is cleaned with an antiseptic solution; and thirdly, a non-adhesive padded dressing is applied to protect and redistribute pressure at the site. At regular follow-up appointments for chiropody, oozing tissue is removed from around the wound to allow epithelization from the edges and for the site to close. After the ulcer has healed, callus is debrided from the site regularly to prevent re-ulceration (*op cit*, **Tindall, 1992**).

For some patients, ulcers may be long-standing because of continued weight-bearing at the site once the ulcer has broken through the skin. In such cases treatment with a below-knee plaster cast (or *total contact cast*) is indicated in preference to the expensive alternative of hospital admission. In cases where the chiropodist allows the patient to walk in the cast, a rubber heel is incorporated into the sole - this type of cast is referred to as a *walking cast*. Plaster casts have been found to be very effective in healing plantar ulcers (**Lang-Stevenson et al, 1985**; **Laing et al, 1991b**) possibly because of the following reasons: (1) the weight of the cast immobilises the patient and the foot is not continually weight-bearing; (2) the infiltration of tissue with fluid (oedema) is controlled and reduced by the support provided by the plaster; (3) the patient is protected from further trauma by the hard plaster shell; and (4) plantar stresses are reduced by the total contact fit.

The hypothesis that plantar stresses are reduced inside plaster casts has been investigated by three groups of workers in two studies of asymptomatic subjects and one study of patients with diabetes. Tests on asymptomatic subjects by **Birke et al (1985)** and **Novick et al (1991)**, showed walking casts to reduce significantly the pressures beneath the forefoot in comparison to shoes. These reductions are attributable to the total contact support of the cast, which distributes pressure beneath the foot to reduce pressure peaks. In the former study of six subjects, four sites were transduced and significant pressure reductions were noted beneath the first and third metatarsal heads. Pressures were also reduced beneath the fifth metatarsal head and heel, but not significantly. Unfortunately the data recorded by these workers was not presented in units of pressure, but instead as millimetre deflections on a chart recorder. As a result, a comparison with pressures from other studies is not possible. The latter study of 20 subjects showed walking casts to reduce significantly the pressures beneath the second and fourth metatarsal heads in comparison to shoes. Pressures were higher beneath the midfoot and only slightly reduced beneath the heel in the walking cast. The increase in pressure beneath the midfoot in the walking cast compared to the shoes is attributable to the total contact support of the medial longitudinal arch and the fact that weight-bearing occurs on the rubber heel beneath this area. The data recorded in this study was presented in pounds force for each of the transduced sites. Converting this data via transducer area to pressure, the average maximum peak pressures for the heel, midfoot and the second and fourth metatarsal heads when wearing shoes were 634 kPa, 232 kPa, 832 kPa and 543 kPa, respectively. In the walking casts, the pressures beneath these same areas were 543 kPa, 311 kPa, 379 kPa and 289 kPa, respectively. In a study of seven patients with diabetes (eight feet), significant reductions in both pressures and longitudinal shear stresses were noted in walking casts compared to conventional leather shoes (**Pollard, 1984**) (table 2.7). Although it could be argued that shear stresses were reduced in the cast because the foot was contained and its mobility was reduced, it is also possible that the reduction was due to a hinderance of the shearing action of the transducer from the surrounding cast as a result of the methodology used.

shoes and orthoses for patients with diabetes

Clinics treating patients with diabetes have different approaches to the provision of special shoes. Some prescribe shoes to patients that are judged at risk of ulcerating, while others only provide shoes once ulceration has occurred and healing has taken place. Ideally, the shoes supplied to patients with diabetes should be made-to-measure and have: (1) low heels to reduce forefoot weight-bearing; (2) lacing to close the quarter and hold the foot back in the shoe; (3) uppers made of a breathable material, such as leather; (4) a high toe box, so that the upper does not rub against the dorsal surface of toes; and (5) adequate depth to accommodate a flat cushioning insole or an insole shaped to the contours of the sole (*moulded insert*) that distributes pressure and reduces pressure peaks (**Tovey, 1984**). Bespoke shoes with the above characteristics are referred to as *custom-made extra-depth shoes*, while those bought *off-the-shelf* with these characteristics are referred to as *stock shoes* (see figure 4.2).

Rocker-bottom or *rocker-soled* shoes are typically supplied to patients with a history of plantar ulceration. The outsole of these shoes is designed with a steep toe clearance that produces an axis beneath the ball about which the shoe can pivot. It is hypothesised that this type of shoe causes the whole foot to tilt forward soon after heel-lift, resulting in push-off before a significant transfer of load to the forefoot and toes. However, the axis about which the shoe pivots is critical in determining the sites beneath the forefoot at which plantar shear and pressure are reduced compared to conventional shoes. In-shoe plantar pressure measurement led **Geary and Klenerman (1986)** to advocate the placement of the rocker axis proximal to the line of the metatarsal heads. However, **Pollard (1984)** found higher pressures beneath the hallux and first metatarsal head in shoes with this rocker design, although longitudinal shear stresses were lower beneath the metatarsal heads compared to conventional shoes; while **Schaff and Cavanagh (1990)** found this rocker design to increase pressures beneath the heel, midfoot and the lateral aspect of the forefoot.

Although cushioning insoles and moulded inserts have been described in the literature as therapeutic aids for the diabetic foot, only a few workers have used objective methods of measurement to assess the pressure attenuating effects offered by such shoe orthoses. Two materials are commonly used in the fabrication of shoe orthoses: PPT (Professional Protective Technology)®[13], a closed-cell foam; and Plastazote®[20] a rigid open-cell foam. For flat cushioning insoles PPT is used; whereas for moulded inserts both PPT and Plastazote are used, typically in a sandwich construction of *low-density Plastazote/PPT/high-density Plastazote* to produce rocker inserts (figure 2.23). The effectiveness of these and other insole materials in reducing peak pressures beneath the foot has been shown by several workers (**Boulton et al, 1984; Pollard, 1984; Brodsky et al, 1988; Smith et al, 1989**). More importantly **Spence and Shields (1968)** showed Neoprene®, a closed-cell foam (the material used to make wet-suits), to be effective in preventing blisters, calluses and ulcers beneath the feet of patients with neuropathy. Unfortunately the shock attenuation of PPT and Plastazote deteriorates in use, the Plastazote more-so than the PPT (**Pratt, 1988**). This deterioration in shock attenuation of the Plastazote is the result of "bottoming-out", ie.

[20] BXL Plastics Ltd, 675 Mitcham Road, Croydon, Surrey, CR9 3AL, UK.

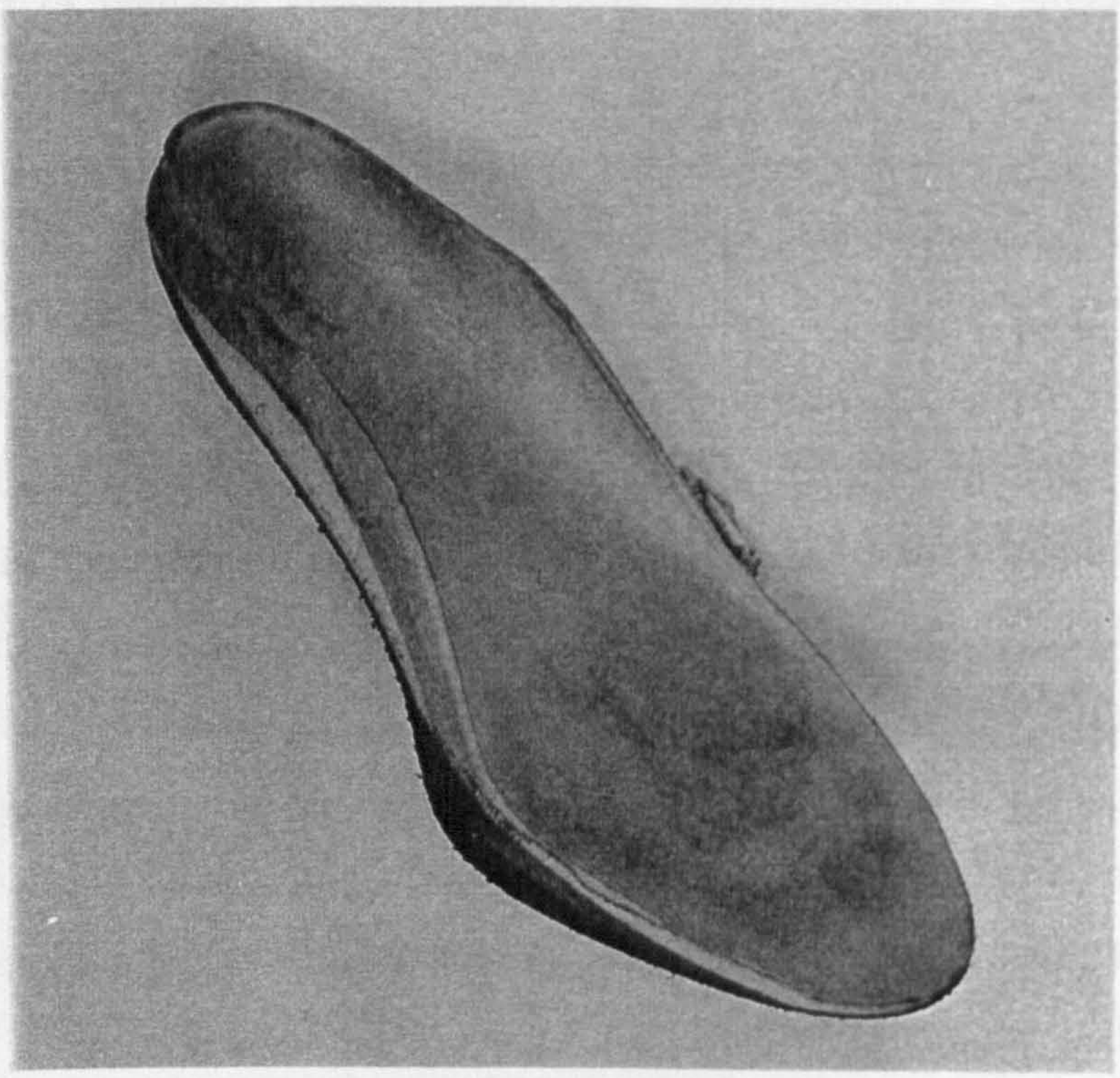

Figure 2.23. Rocker insert of a sandwich construction: low-density Plastazote - top; PPT - middle; and high-density Plastazote - bottom. The top is covered with leather.

compression with a consequent increase in stiffness, under repetitive load or long term (creeping) loading, which has been observed in laboratory tests (**Campbell et al, 1984; Brodsky et al**). Bottoming-out of Plastazote reduces its cushioning effect beneath areas of high pressure, which can be detrimental to the insensitive foot.

At present only one group of workers has investigated the effectiveness of moulded inserts in reducing peak pressures beneath the forefoot (**Lord and Hosein, 1994**). The study of six patients with diabetes measured in-shoe plantar pressures using the insole sensor developed by the TekScan[8] company. A significant reduction in the local maximum peak pressures beneath the medial side of the forefoot was found with the use of moulded inserts compared to a flat inlay. This pressure reduction was attributed to simultaneous pressure redistribution into the midfoot. However, it is apparent that at the time of maximal forefoot loading the medial longitudinal arch is not weight-bearing. Instead, the reduction in local maximum peak pressures beneath the medial side of the forefoot may be due to the insert altering the biomechanics of the foot through the support provided into the medial longitudinal arch. Subtalar eversion may be reduced during forefoot loading and hence there may be a subsequent reduction in the medial transfer of load across the forefoot.

CHAPTER 3
INSTRUMENTATION FOR SHEAR STRESS MEASUREMENT

3.1 INTRODUCTION

The data gathered from in-shoe plantar stress measurements directly permits two important investigations: that of shoe design modifications on shod foot function and the effectiveness of footwear management on plantar pathologies. In the last decade there has been a growing trend towards the development of high resolution pressure sensitive insoles, whose use has provided comprehensive plantar pressure data. In comparison, very few research groups have developed devices for shear measurement, primarily because of a lack of a suitable technology.

The few transducers that are now available for shear measurement are discrete and are restricted to making measurements along one direction, ie. they are *uni-axial.* Workers using these transducers have all employed static methods of calibration prior to making dynamic measurements having apparently overlooked the relevance of dynamic calibration.

Discrete shear transducers require precise placement at anatomically referenced sites beneath the foot for accurately defined measurements. This can be a time-consuming process, so an insole sensor with a high spatial resolution would be advantageous. The current solution to the design of such a sensor has been to instrument an inlay, ie. flush-mount several discrete transducers into a custom-made insole. At present, the number of shear transducers that can be mounted side-by-side into an inlay is limited by the sizes of the available devices.

Although the available shear transducers are uni-axial, they are quite often used to make measurements in orthogonal directions at a single plantar site. This requires successive walks in which the axis of the transducer is first orientated in one direction and then the other. Because successive walks may differ considerably in many respects, a bi-axial shear transducer would be advantageous for these measurements. A transducer was recently developed at King's College, London, for measuring two orthogonal directions of shear, and pressure, at the interface between the residual stump and prosthetic socket of patients with a below-knee amputation (**Williams, 1993**). By making adaptations to this tri-axial transducer, it has been possible to measure orthogonal directions of plantar shear in-shoe. This Chapter describes this shear transducer, its electronic interfacing, data collection system and method of quasi-static calibration.

3.2 SYSTEM OVERVIEW

The various units of the data acquisition hardware are shown in figure 3.1. The shear transducers are interfaced to a unit which is worn as a backpack which contains voltage excitation circuits and signal pre-amplifiers. Ten metres of ribbon cable convey the amplified signals to a second unit which provides low pass filtering. The conditioned signals are then interfaced to a PC for storage. The timing of gait events, particularly heel-contact and push-off, is provided by the inclusion of two footswitches in the system.

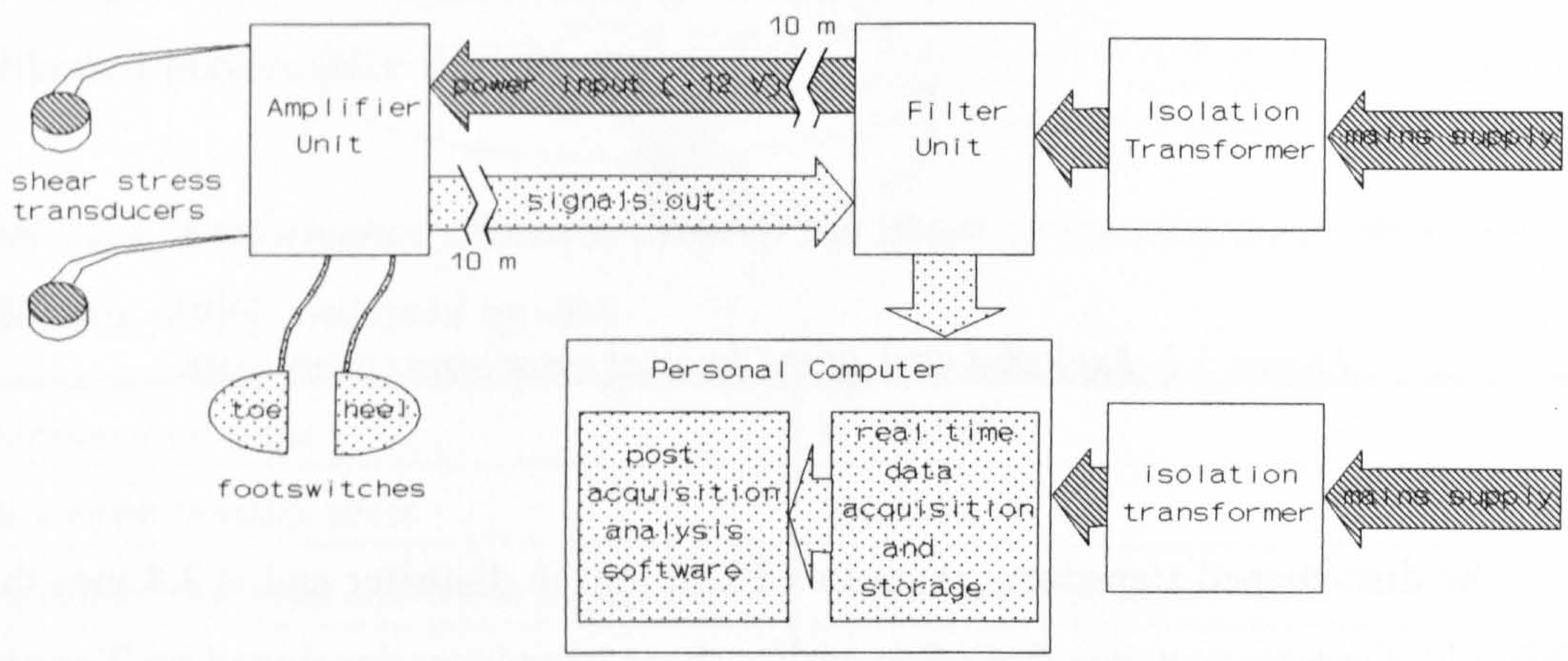

Figure 3.1. Block diagram of the shear transducer data acquisition hardware.

3.2.1 transducer for bi-axial shear stress measurement

The measurement of shear requires the registration of relative movement or the tendency to relative movement between two objects. There may be a *tendency* to relative movement between two objects, ie. with no *actual* movement, if the frictional contact between the objects is sufficiently high to prevent slip.

The principle of operation of the shear transducer is based on the interaction of a magnet and electrically excited magneto-resistor. The voltage output of the magneto-resistor changes in proportion to the overhead displacement of the magnet.

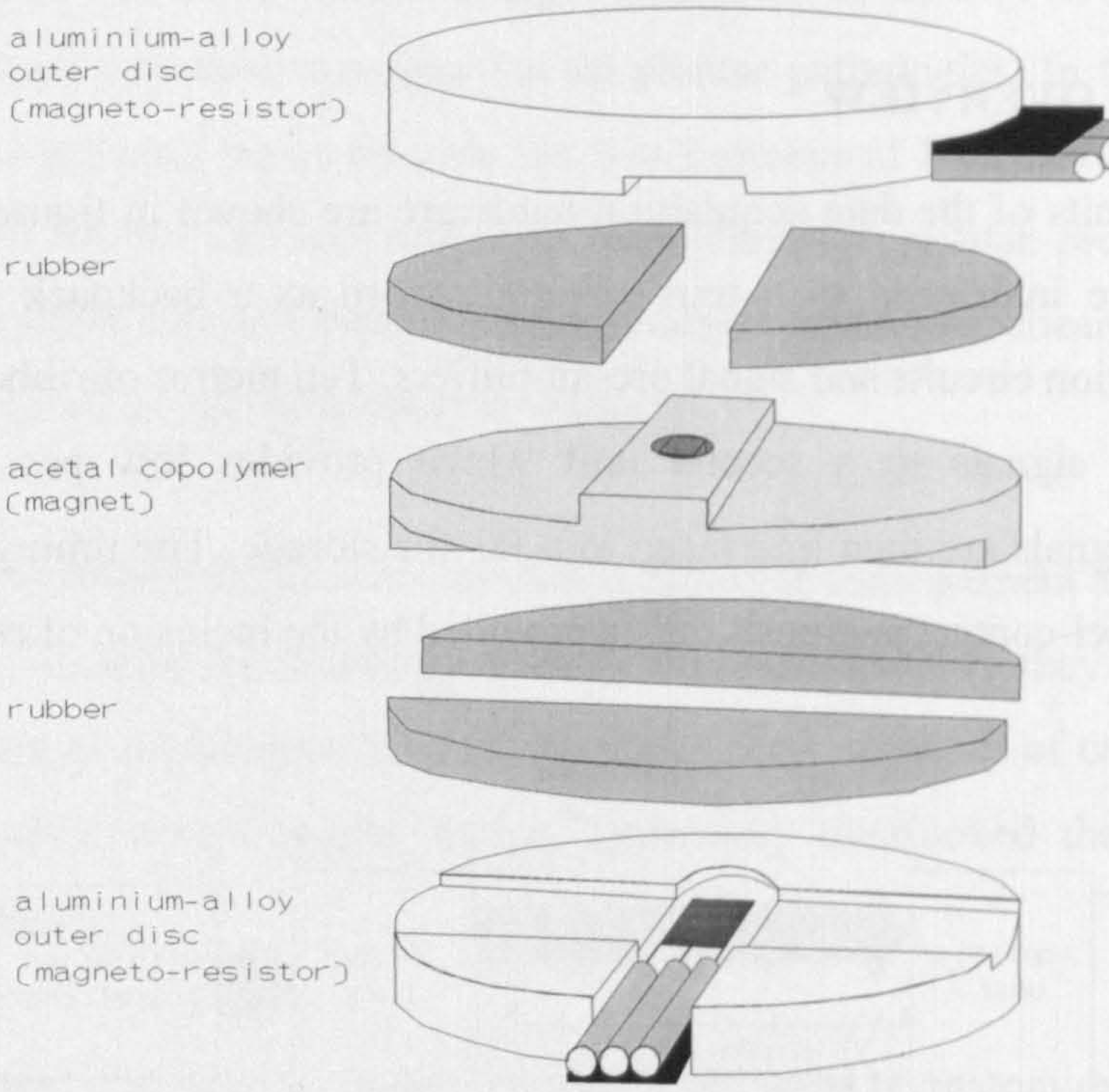

Figure 3.2. Exploded view of the bi-axial shear stress transducer.

The disc-shaped transducer measures 15.96 mm in diameter and is 3.8 mm thick. This odd diameter matches that of an earlier shear transducer developed by **Tappin et al (1980)**, who chose this value to produce a device with a surface area of exactly 2 cm^2. Their reason for doing this was to speed up the process of converting the transducer voltage output to stress. The size of the transducer is also comparable to previous discrete devices for pressure measurement and is suitable for recording data beneath a single metatarsal head. The transducer consists of three discs separated by natural rubber (figure 3.2). Two outer aluminium alloy discs each house a magneto-resistor and have a groove. The central acetal copolymer disc houses a centrally placed

Samarium Cobalt magnet and has a ridge on either side, which are orthogonal to each other. Displacement of the discs is restricted to two orthogonal directions by the interaction of the ridges and grooves on the opposing disc surfaces. Each orthogonal direction of movement is referred to as an *axis*; one is assigned the *x-axis* and the other the *y-axis*. The rubber provides a resistive force to shearing of the transducer and restores the transducer to its equilibrium position when shearing forces are removed.

A comprehensive calibration was performed on a batch of ten of these transducers by **Williams (1993)**. This provided the performance characteristics listed in table 3.1. **Williams** specified *accuracy* as the error incurred when using the gradient of the first order polynomial curve fitted to the calibration data as the conversion constant for processing the clinical data. The accuracy was defined to include errors due to both hysteresis and non-linearity and was found by plotting the absolute maximum value between the hysteresis loop and the first order polynomial curve to each of the calibration graphs for load cycles of ±5 N, ±10 N, ±20 N and ±50 N. The *bandwidth* specifies the range of frequencies over which the gain (the ratio of output:input) is constant to within -3 dB. During development, four transducers were subjected to more than one-hundred-thousand loading cycles (±50 N, 1 Hz) without degradation of their calibrated performance (*op cit*).

*Table 3.1. Performance characteristics of the shear stress transducer developed by **Williams** (1993) (adapted op cit).*

Measurement range	0 - 250 kPa
Maximum tolerable stress	2500 kPa
Accuracy (first order conversion polynomial)	±9% of indicated stress
Bandwidth	d.c. - 750 Hz
Temperature range	20°C - 40°C
Life	>100,000 gait cycles

3.2.2 transducer interfacing electronics

Each shear transducer was interfaced to electronic circuits which provided voltage excitation and signal conditioning. For each shear axis, the magneto-resistor formed two arms of a Wheatstone bridge circuit. A voltage excitation circuit provided a constant 1 V supply across the two arms of the magneto-resistor and the output signal was taken between the centre-tapped termination of the magneto-resistor and analogue ground. For an applied shear force of ±40 N, the transducer output signal is typically ±100 mV. A differential amplifier with a gain of 40 was used to map this signal to the ±5 V input range of an A/D converter.

Ten metres of twenty-way ribbon cable conveyed the amplified transducer signals to a filter unit. Four-pole Bessel filters were used to exclude frequency content above 100 Hz, ie. above the frequency of interest. The filter circuit used here had a lower cut-off frequency than the circuit used by **Williams (1993)**. With an expected signal bandwidth of less than 50 Hz, **Williams** had designed his filter circuit with a 250 Hz cut-off to provide a constant group delay over the passband, with a less than 1% deviation in amplitude from d.c to 50 Hz. However, with this circuit it was possible that noise signals with frequencies between 100 Hz and 250 Hz, outside the range of interest, could corrupt the data being recorded. The circuit diagram for the redesigned filter appears in figure 3.3.

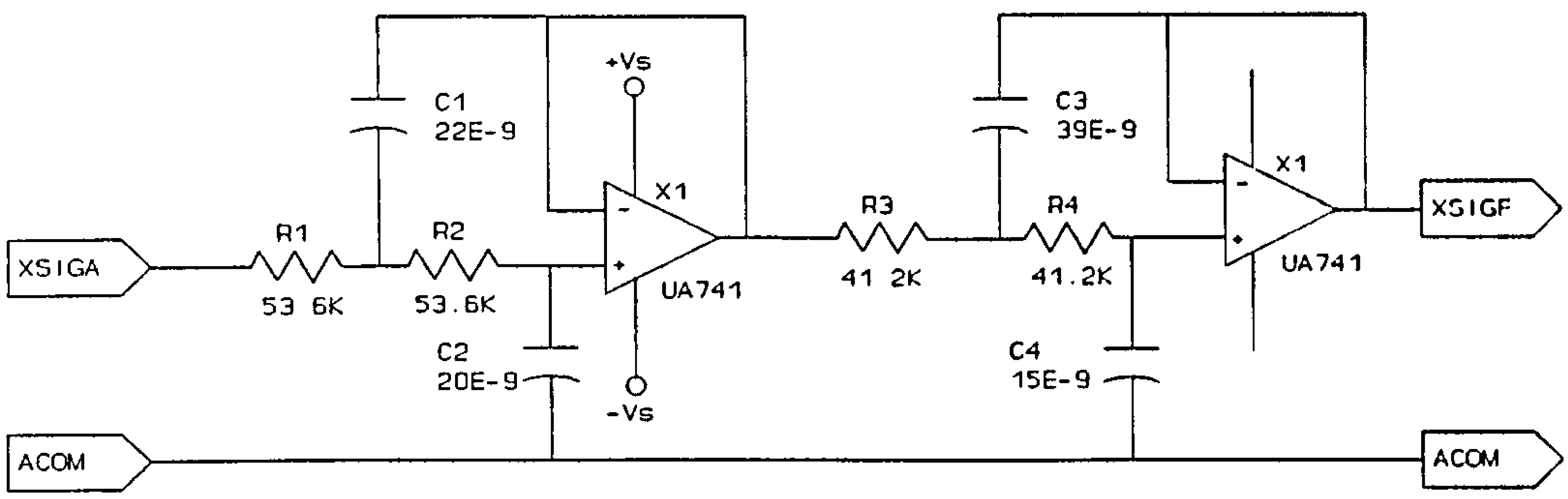

Figure 3.3. Circuit diagram of the redesigned filter used in the shear transducer signal-conditioning electronics: 4-pole Bessel filter with a 100 Hz cut-off frequency.

3.2.3 analogue-to-digital conversion

The transduced signals were interfaced to a personal computer (Viglen PC AT) using a Data Translation[21] *DT2814* A/D converter. The DT2814 provides sixteen single-

[21] Data Translation Ltd., The Mulberry Business Park, Wokingham, Berkshire, RG11 2QJ, UK.

ended input channels; 12-bit resolution; selectable analogue input ranges of 0-5 V, ±2.5 V and ±5 V; and allows a data throughput of 25 kHz. Eight of the sixteen channels were used to sample shoe-to-ground contact data from two footswitches[22] (type FSW12 for the toe and FSW7 for the heel) and shear data from three bi-axial transducers simultaneously. The ±5 V input range was used throughout.

3.3 DATA ACQUISITION SOFTWARE

During data collection the A/D converter was controlled by *Labtech Notebook* (version 7), a software package produced by Laboratory Technologies Inc[23]. A sampling frequency of 400 Hz per channel was chosen, ie. four times the maximum frequency of interest. This gave a data throughput (aggregate sampling rate) of 3200 Hz, which was in fact the highest possible because of the available computer RAM. The necessary factors to consider when selecting the sampling frequency are discussed in detail in Chapter 2, section 2.5.3.

3.4 SYSTEM INTEGRATION AND PERFORMANCE

A total of twelve voltage excitation and pre-amplifier circuits for interfacing six shear transducers were contained in a 300 x 250 x 50 mm^3 unit weighing approximately 1.5 N, which was worn as a backpack (figure 4.9). The filter unit, which also contained circuits for interfacing six shear transducers, was sited next to the computer. The power for both these units was provided by a ±12 V bench supply; and the 10 m signal cable between them was taped at intervals along its length to the power supply cable for the backpack. A force of approximately 2.5 N was required to drag these two cables across the linoleum floor of the laboratory where measurements of in-shoe plantar shear were made.

The 12 V supply for the footswitches was derived from the bench power supply. A pressure of approximately 5 kPa is required to close this footswitch, which then generates an output voltage of approximately 0.7 V.

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22 MIE Medical Research Ltd., 6 Wortley Moore Road, Leeds, LS12 4JF, UK.

23 Laboratory Technologies Corporation, 400 Research Drive, Wilmington, MA 01887, USA.

Prior to safety tests, signal continuity was checked, along with the amplifier gains, filter cut-off frequencies, power supply voltages and the transducer supply voltages. A 20 Hz, 40 mV_{p-p} input signal was used to check the amplifier gains. This was 40 for all except one of the transducer interface circuits, which had a gain of 42.5 for both axes. The -3 dB cut-off frequencies of the filters ranged from 115 Hz to 123 Hz.

The magnitude of the noise signal was determined with all components of the system connected and the ribbon cable trailed around the floor of an electronics laboratory. Data was digitized at 200 Hz for 20 seconds, while the ribbon cable was shaken to simulate its movement during walking. During this test three transducers were interfaced. The three spare transducer interface circuits were loaded using "dummy" transducers, which simulate additional transducers using 330 Ω resistors for each arm of the magneto-resistor. The magnitude of the noise signal on each of the transducer axes did not exceed ±50 μV. For the maximum expected transducer output signal of ±100 mV, for an applied load of ±40 N (see section 3.2.2) this gives a signal-to-noise ratio of 66 dB.

3.5 ELECTRICAL SAFETY

The PC had its own mains power supply: the pre-amplifier and filter units were powered by a ±12 V bench supply. To reduce earth leakage currents in the system, which may be a potential source of shock to the subject/patient, the mains supply to the PC and the bench supply was derived through isolation transformers. These provided a "floating" supply, which reduced earth leakage currents through not being referenced to earth unlike the mains supply. In addition the chassis of the pre-amplifier backpack was connected to earth.

Prior to use the entire system was safety tested by the Safety Officer in the Department of Medical Engineering and Physics, according to the electrical safety standard set down in BS5724[24].

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[24] BS5724 - safety requirements for Medical Electrical Equipment.

3.6 TRANSDUCER CALIBRATION PROCEDURE

The response of the transducers to cyclic loading was determined before *and* after the clinical trial. The post-trial calibration was used to check the constancy of the transducers' response after an application. The quasi-static calibration that was performed used the equipment and followed the procedure outlined by **Williams (1993)**. A brief description of the calibration equipment and procedure is given here.

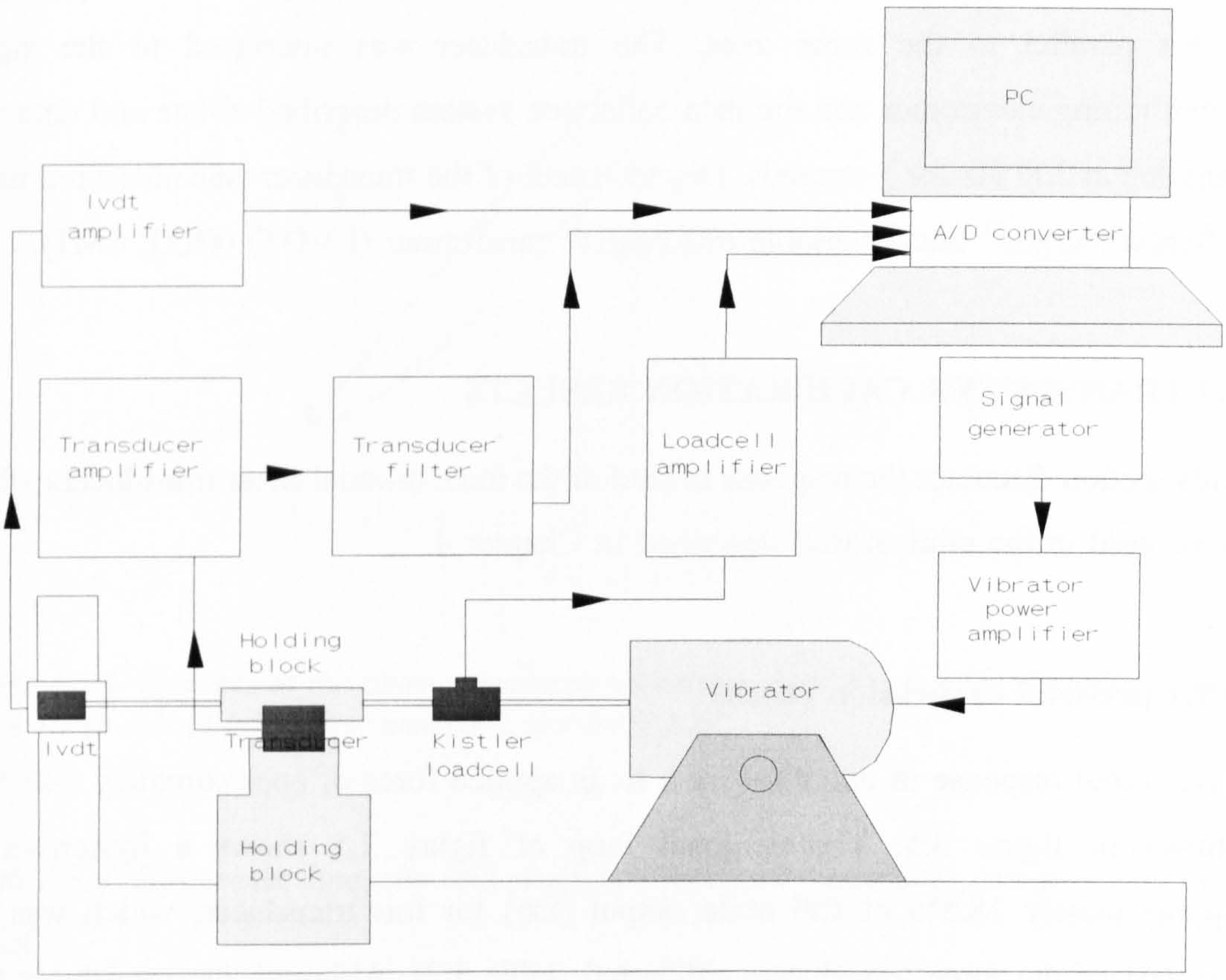

Figure 3.4. Block diagram of the shear transducer calibration system.

The transducer was held horizontally between two cylindrical aluminium alloy blocks in the calibration rig (figure 3.4). The lower block held one of the outer discs of the transducer stationary, while the upper block held the other outer disc and was located in a sliding block running on linear bearings. The sliding block was linked to an industrial vibrator[25] (Gearing & Watson GWV100 A) which was controlled by a power amplifier (Gearing & Watson SS300). The frequency source for the power amplifier was provided by a Global Specialties signal generator (series 8120)[26]. Forces applied by the vibrator were measured by a Kistler[27] loadcell (type 9301A), which was interposed between the linking arm and the sliding block. The loadcell was interfaced to a Kistler charge amplifier (type 5007).

During calibration, a force of approximately ±40 N was applied at a frequency of 1 Hz parallel to the shear axes. The transducer was interfaced to the signal conditioning electronics and the data collection system described above and data was sampled at 200 Hz for 5 seconds. Displacement of the transducer was measured using a Schlumberger[28] linear variable differential transformer (LVDT) (OD3, SM1).

3.7 TRANSDUCER CALIBRATION RESULTS

This section discusses the response to load of the three bi-axial shear transducers which were used in the clinical trial described in Chapter 4.

3.7.1 pre-trial calibration results

The output response of one transducer to an applied force of approximately ±40 N is shown in figure 3.5. The diagonal loop of figure 3.5 shows a hysteresis of approximately 18.5% of full scale output (fso) for this transducer, which was the greatest of the three transducers calibrated (table 3.2). Although the transducer was expected to exhibit hysteresis because of the presence of rubber in the construction, the loop was expected to be uniform in width. Energy losses in rubber tend to be manifest by a hysteresis loop of uniform width, so this would have confirmed that the energy losses were in the rubber. The central portion of the loop suggests that the energy losses over this range of the load cycle *were* predominantly in the rubber. The

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25 Gearing & Watson (Electronics) Ltd., South Road, Hailsham, East Sussex, BN27 3JJ, UK.

26 Tabor Electronics Ltd, 3 Hamlacha Street, P.O.B. 901, Haifa, Israel 31008.

27 Kistler Instrument AG, CH-8408 Winterthur, Switzerland.

28 Schlumberger (UK) Ltd., Ferndown Industrial Estate, Wimborne, Dorset, England.

narrowing of the hysteresis loop at the extremes of the load cycle was probably the result of localised friction between the ridge and groove of the transducer discs at the extremes of their movement.

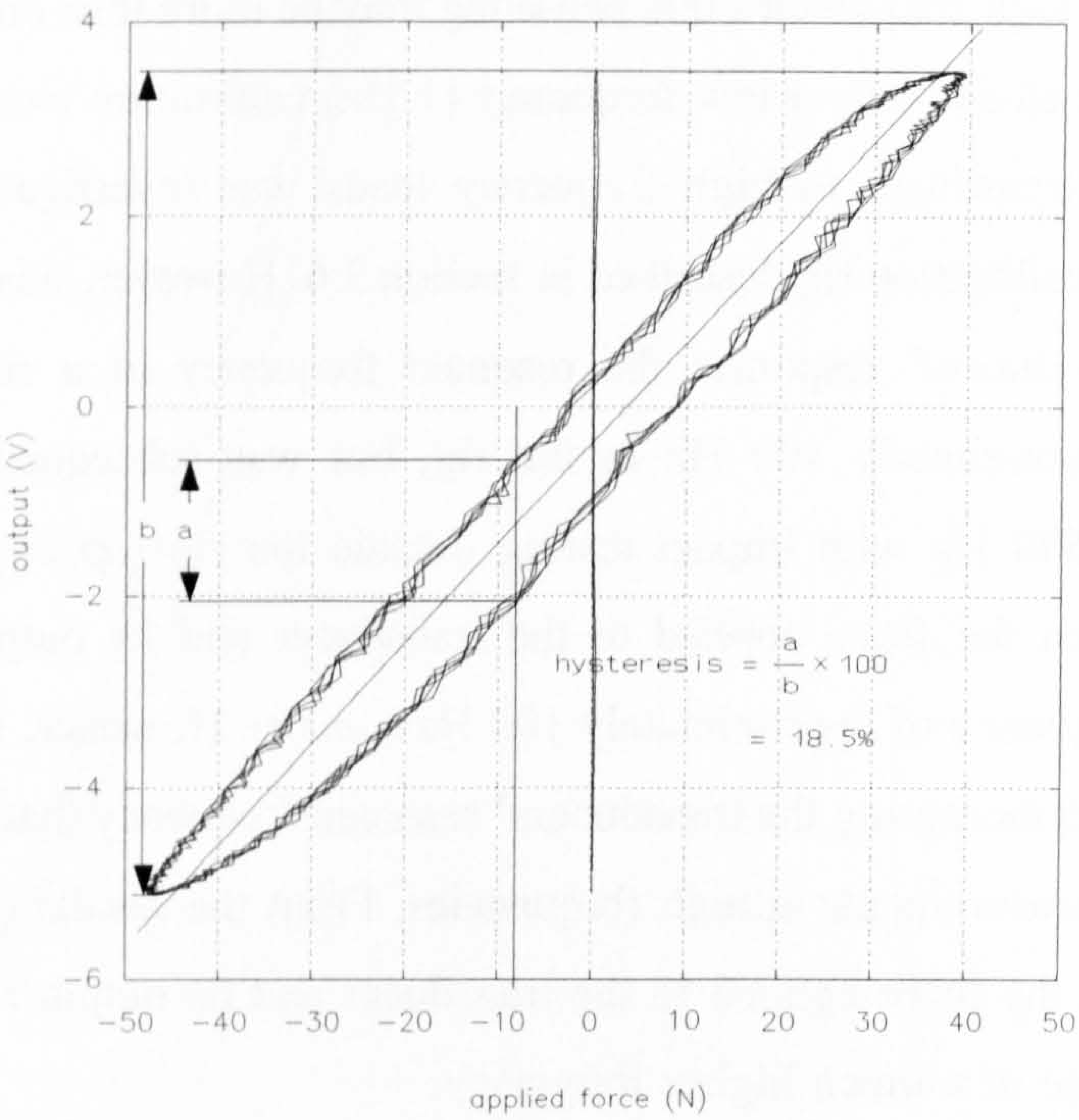

Figure 3.5. Example of the shear transducer response to load for quasi-static calibration at 1 Hz (x-axis of transducer number 1, see table 3.2).

Table 3.2. Hysteresis, linearity and cross-talk for each transducer axis pre- and post-trial.

Transducer axis	Pre-trial calibration			Post-trial calibration		
	hysteresis (% fso)	linearity (% ±30 N)	cross-talk (% fsl)	hysteresis (% fso)	linearity (% ±30 N)	cross-talk (% fsl)
1x	18.5	2.2	2.5	13.8	1.9	3.9
1y	12.2	0.9	2.5	10.4	2.1	4.3
2x	15.7	0.9	3.9	9.7	0.9	4.2
2y	16.3	1.4	5.9	8.1	0.8	6.7
3x	18.0	1.4	1.9	13.2	9.1	5.7
3y	14.9	3.1	2.5	17.1	4.6	2.3

The rubber in the transducer assembly may also contribute to a time delay between the force applied to the transducer and its output response, ie. a *phase lag*. The phase lag in the response of rubber is frequency dependent: as the frequency of the applied force increases, the time delay between the applied force and the output response will increase. At very high frequencies this phase lag may be more than one cycle, ie. more than 360^0. In this study, only a low frequency (1 Hz) calibration was performed. The response of the transducer to high frequency loads was investigated by **Williams (1993)** using the calibration rig described in section 3.6. However, this rig significantly damped the transducers' response: the resonant frequency of a shear section was measured at approximately 100 Hz in the rig, but was subsequently found to be approximately 1500 Hz with impact testing outside the rig (*op cit*). Evidence of a phase lag between the force applied to the transducer and its output response was apparent at a frequency of approximately 100 Hz (*op cit*). However, it is evident from the discrepancy in measuring the transducers' resonant frequency that the rig could not make accurate measurements at high frequencies. From the results of the impact test it is thought that the force applied to the transducer and its output response begin to move out of phase at a much higher frequency.

Linearity was calculated as the difference between the first and fifth order polynomial curves to the calibration data over the range ±30 N. The greatest linearity error noted for pre-trial calibrations was approximately 3%. This was for the y-axis of transducer number 3 (figure 3.6).

To check for *mechanical cross-talk*, the output from the other axis of the transducer was recorded simultaneously during calibration. This is the vertical hysteresis loop in figure 3.5, which should ideally be zero. The output that was recorded was the result of minute movement between the walls of the ridge of the central acetal copolymer disc and the groove of the outer aluminium alloy disc toward and away from each other. The greatest cross-talk noted during pre-trial calibrations was approximately 6% of full scale load (fsl) for the y-axis calibration of transducer number 2 (table 3.2).

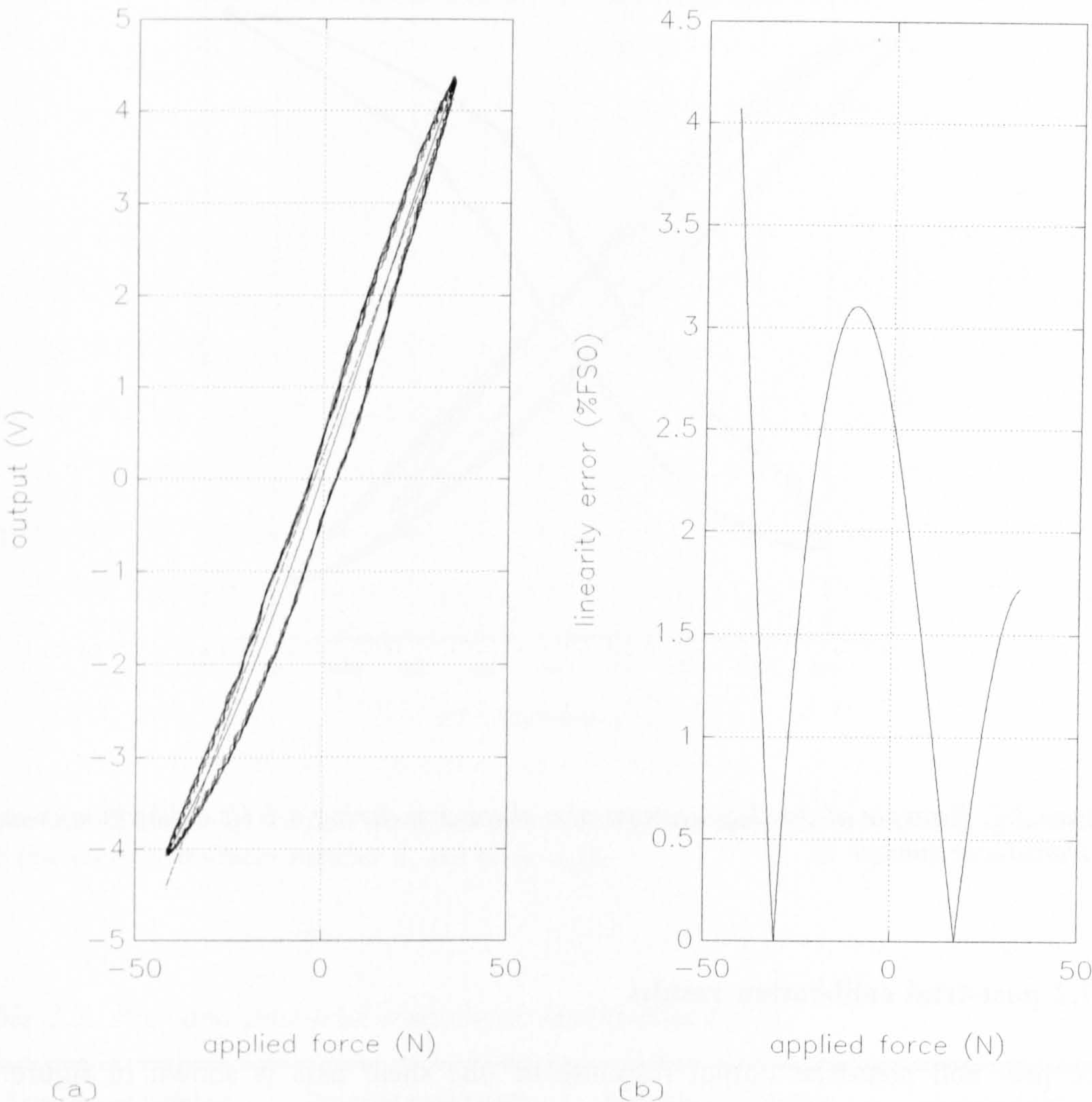

Figure 3.6. (a) Pre-trial response to load for the y-axis of transducer number 3. First and fifth order polynomial curves shown. (b) Linearity error for the y-axis of transducer number 3, ie. the difference between the first and fifth order polynomial curves.

An example of the displacement of one shear axis during calibration is presented in figure 3.7. For a force of approximately ±40 N there was a total displacement of approximately 1 mm on this axis. It was expected that the shape of the graphs of *applied force vs transducer output* and *applied force vs transducer displacement* would be identical for each transducer axis calibrated, however, differences were noted in each case (eg. compare figures 3.5 and 3.7). The graphs of applied force vs transducer displacement are thought to be inaccurate, after close inspection of the LVDT revealed the core was distorted.

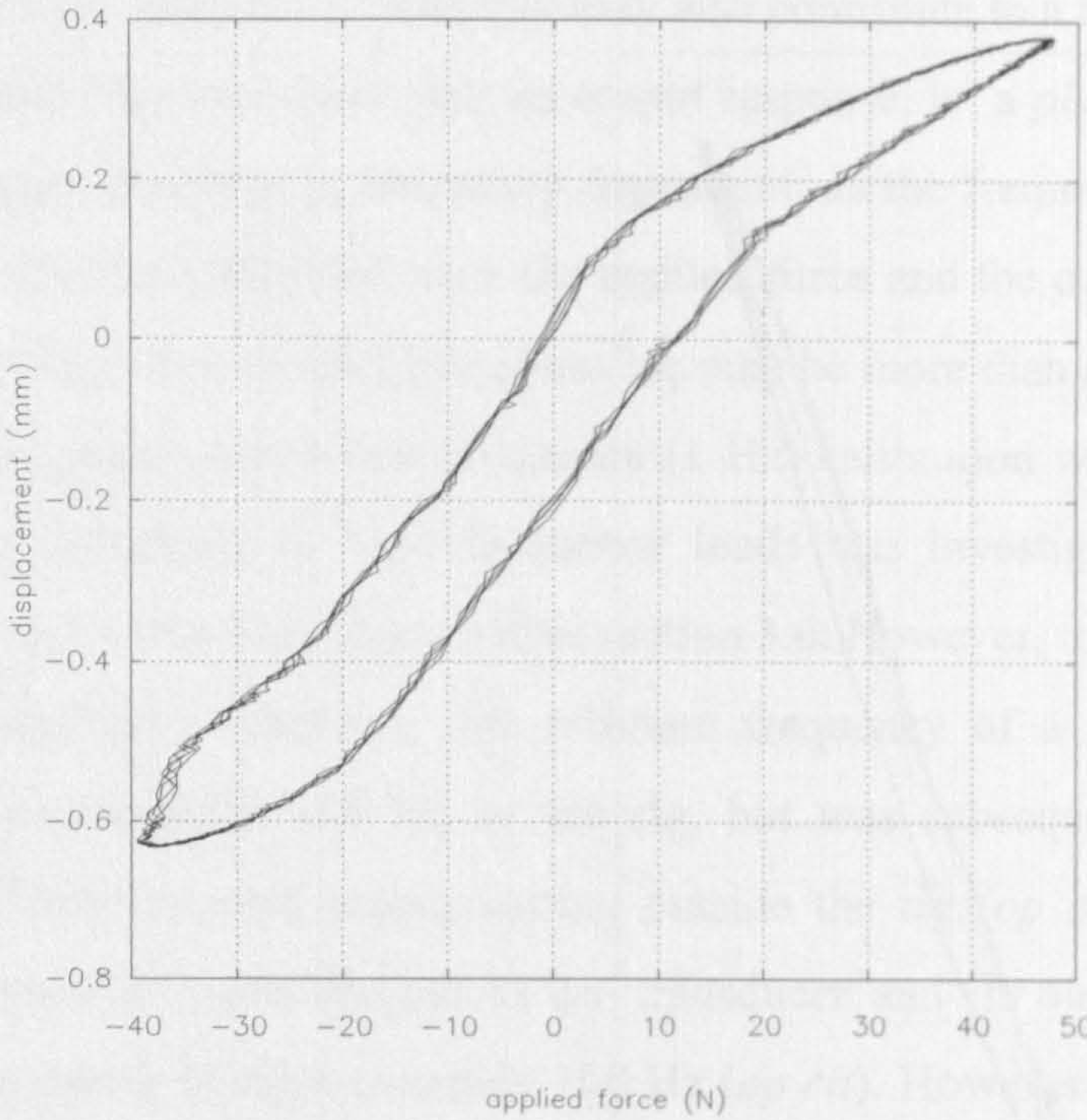

Figure 3.7. Example of the displacement of a shear axis during a 1 Hz calibration (x-axis of transducer number 1).

3.7.2 post-trial calibration results

The pre- and post-trial output response of one shear axis is shown in figure 3.8. Hysteresis was generally found to be lower for each transducer axis in post-trial calibrations compared to pre-trial calibrations; while linearity and cross-talk were generally found to be higher (table 3.2).

The sensitivity of *each* transducer axis was comparable in pre- and post-trial calibrations. Table 3.3 presents the pre- and post-trial transducer sensitivities, ie. the gradients of the first order fit to the pre- and post-trial calibration graphs. For each axis, the gradients to the pre- and post-trial calibration graphs were averaged and used to convert the clinical data from voltage to force.

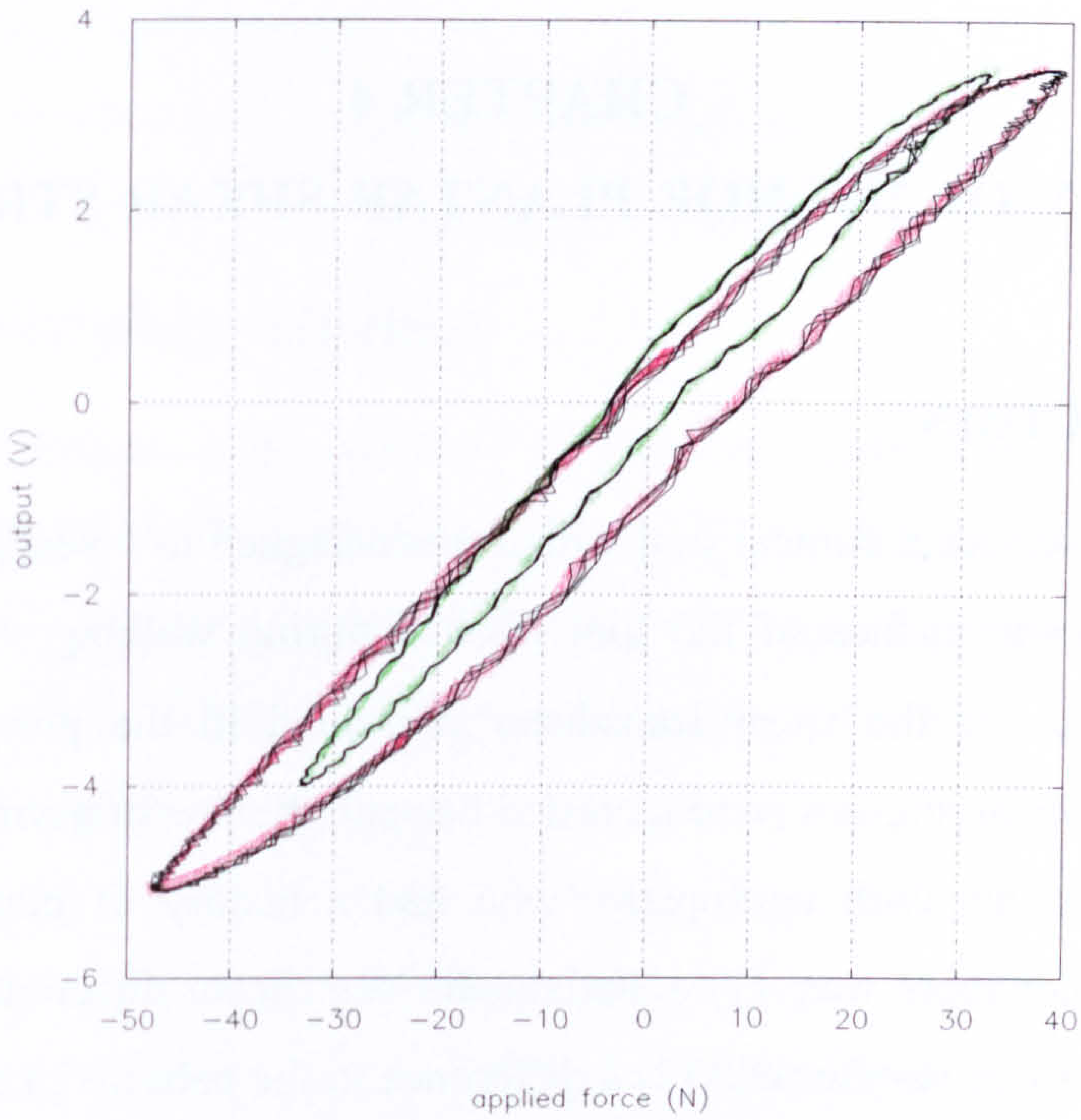

Figure 3.8. Example of the pre- (red) and post-trial (green) calibration response of one shear axis (x-axis of transducer number 1, see table 3.2).

Table 3.3. Pre- and post-trial transducer sensitivities (N/V).

Transducer number & axis	Pre-trial calibration constants	Post-trial calibration constants	Average (sd)
1, x-axis	9.5	8.5	9.0 (0.7)
1, y-axis	11.8	12.1	12.0 (0.2)
2, x-axis	10.4	9.2	9.8 (0.8)
2, y-axis	12.5	12.0	12.3 (0.4)
3, x-axis	11.4	11.6	11.5 (0.1)
3, y-axis	8.8	8.6	8.7 (0.1)

CHAPTER 4
STUDY OF IN-SHOE PLANTAR SHEAR STRESSES

4.1 INTRODUCTION

This Chapter describes a clinical trial which was designed to investigate shear stresses beneath the plantar surface of the foot in-shoe during walking. A methodology is described for use of the shear transducer presented in the previous Chapter for measurements. Shear stresses were recorded beneath the feet of asymptomatic subjects and diabetic patients with neuropathy who had a history of plantar ulceration to determine whether there was: (1) a statistically significant difference in the resultant maximum peak shear stresses and (2) a difference in the patterns of shear, between the two groups. Foot mobility, plantar pathologies and peripheral sensation in the foot were assessed to determine their relationships to the plantar shear stresses measured.

4.2 AIMS AND OBJECTIVES

The aim of this study is to investigate shod foot function based on the measurement of in-shoe plantar shear stresses. The specific objectives of this study are:

- to develop a methodology for measuring in-shoe plantar shear stresses;
- to describe normal shod foot function based on the measurement of in-shoe plantar shear stresses and the assessment of joint ranges of motion;

and

- to determine the differences, if any, between the plantar shear stresses of asymptomatic subjects compared to diabetic patients with neuropathy and a history of plantar ulceration.

4.3 TRIAL PROCEDURE

The clinical trial was conducted in the following stages:

- Subject and patient recruitment.
- Pressure measurement.

- Shear stress measurement.
- Biomechanical examination.
- Neurological assessment of patients with diabetes.

4.3.1 subject and patient recruitment

Asymptomatic subjects were recruited to the trial from staff members of two departments within King's College Hospital, London: the Department of Medical Engineering & Physics and the Diabetic Foot Clinic. Age, body mass and sex were not criteria for exclusion. Instead selection was made on the basis that as many staff members of the two departments could fit comfortably into one of two available pairs of extra-depth stock orthopaedic shoes[29], either a UK size 7½ or 9 (figure 4.2). The shoes were 3-hole Gibsons with a 19 mm heel. Standard footwear of this type was required to accommodate the instrumentation used for shear measurement; and the sizes were chosen to provide a comfortable fit to as many of the staff members of the two departments.

Patients were recruited from those routinely attending the Diabetic Foot Clinic at King's College Hospital. Age, body mass and sex were again not criteria for exclusion; and the patients were not matched to the subjects. Consecutive patients were selected on the basis that they:

- wore custom-made orthopaedic shoes with moulded inserts;
- had peripheral neuropathy;
- had at least one foot free of gross deformity, ie. Charcot deformity and ray amputations;
- had healed plantar metatarsal ulcers;
- did not have active plantar ulcers beneath the forefoot requiring padded dressings;
- did not have ischaemia;

and

- were able to walk unaided.

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[29] LSB Orthopaedic Ltd., 309 Sovereign Road, King's Norton Business Park, Birmingham B30 3HU, UK.

Peripheral neuropathy and ischaemia were assessed by routine clinical examination, the results of which were made available.

4.3.2 pressure measurement

Plantar pressure was recorded to locate the sites of peak pressure beneath the metatarsal heads and heel so that the shear transducers could be located in regions of maximum load associated with these anatomical sites (**Lord et al, 1992**). However, this was not the limit of the analysis performed on the pressure data, which was in fact as comprehensive as the analysis of the shear data.

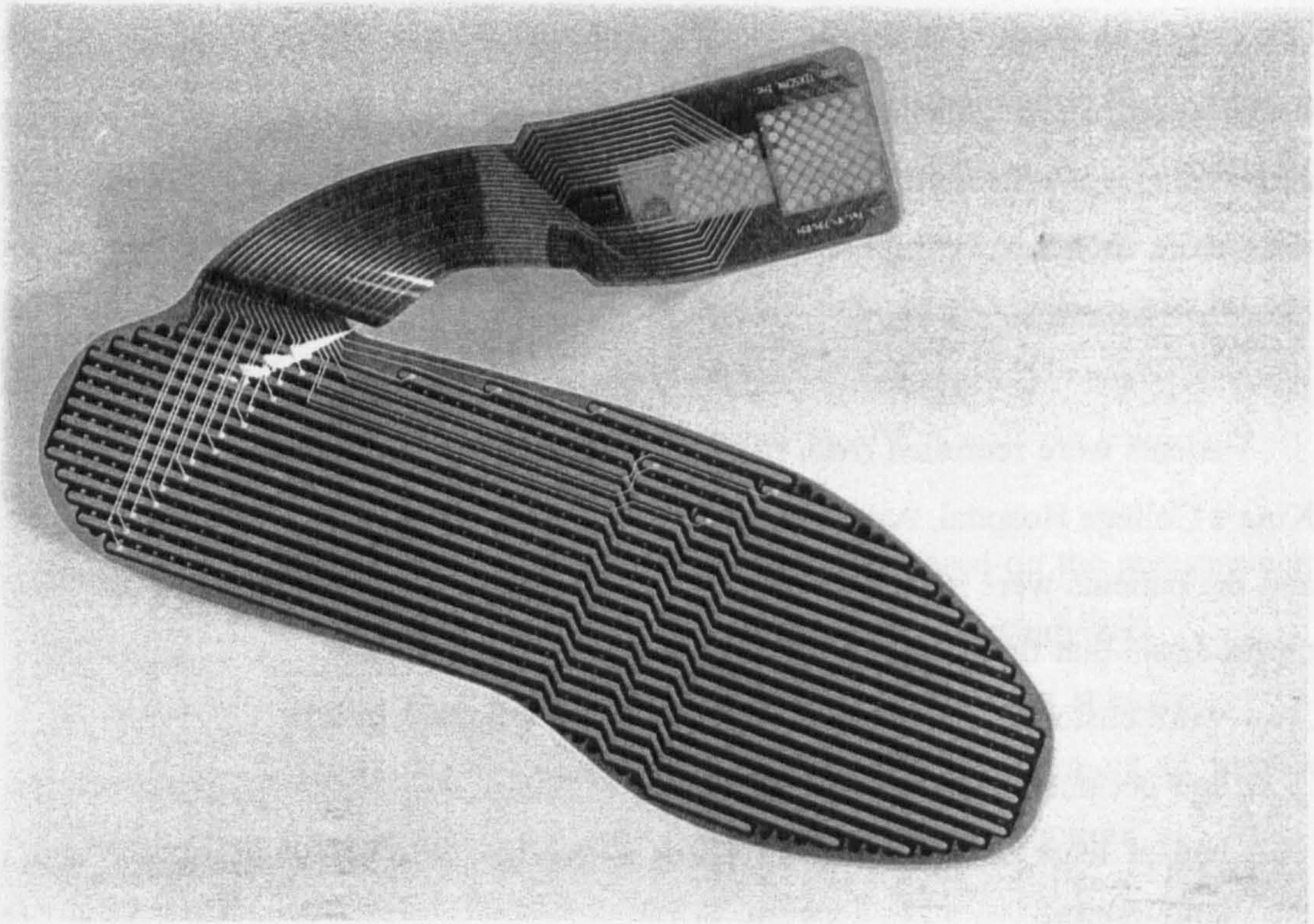

Figure 4.1. The pressure sensitive insole supplied with the F-Scan Gait Analysis System.

Pressure was recorded in-shoe using the *F-Scan Gait Analysis System*, version 3.421, supplied by *TekScan Inc*[8]. This system provides a 0.2 mm thick, flexible pressure-sensitive insole in a standard UK size 12 (US 14, Continental 47) (figure 4.1). The sensors are based on force-sensitive resistive (FSR) technology. Each full size insole has 960 loadcells spaced 5 mm apart, which are formed by the intersection of 3.5 mm wide conductive polymer tracks deposited along and across individual halves of a Mylar film. The insole can be trimmed with scissors to fit a range of shoe sizes down to a UK size 2 (US 3, Continental 35). A handle extending from the side of the insole interfaces to a cuff unit which provides signal conditioning. The cuff unit is worn above the ankle and interfaces to a PC via a 9.5 m length of coaxial cable (figure 4.2). Data collection is enabled from each loadcell at a selectable frequency within the range 1 Hz to 100 Hz.

Figure 4.2. Subject instrumented with the F-Scan Gait Analysis System. Pressures were recorded beneath both feet simultaneously.

Data on the pressure threshold of the insole sensors was not supplied by the TekScan company, but this was investigated in a simple experiment by applying a force over an area of the sensor. A rectangle of Poron®, a polyurethane foam sheet, was placed on the sensor and an aluminium-alloy plate was placed on top. By standing on the plate, bodyweight (approximately 650 N) was applied over the area of the Poron. The pressure applied to the sensor was varied by using different areas of the foam sheet. At 30 kPa applied pressure all the cells were registering and the average transduced pressure over a central area of the sensor registered correctly (31 kPa). At 20 kPa applied pressure, approximately 40% of the loadcells were not registering, which indicated the applied pressure had dropped below the threshold of the sensor (figure 4.3). It was concluded that the pressure threshold of the F-Scan sensor was between 20 kPa and 30 kPa.

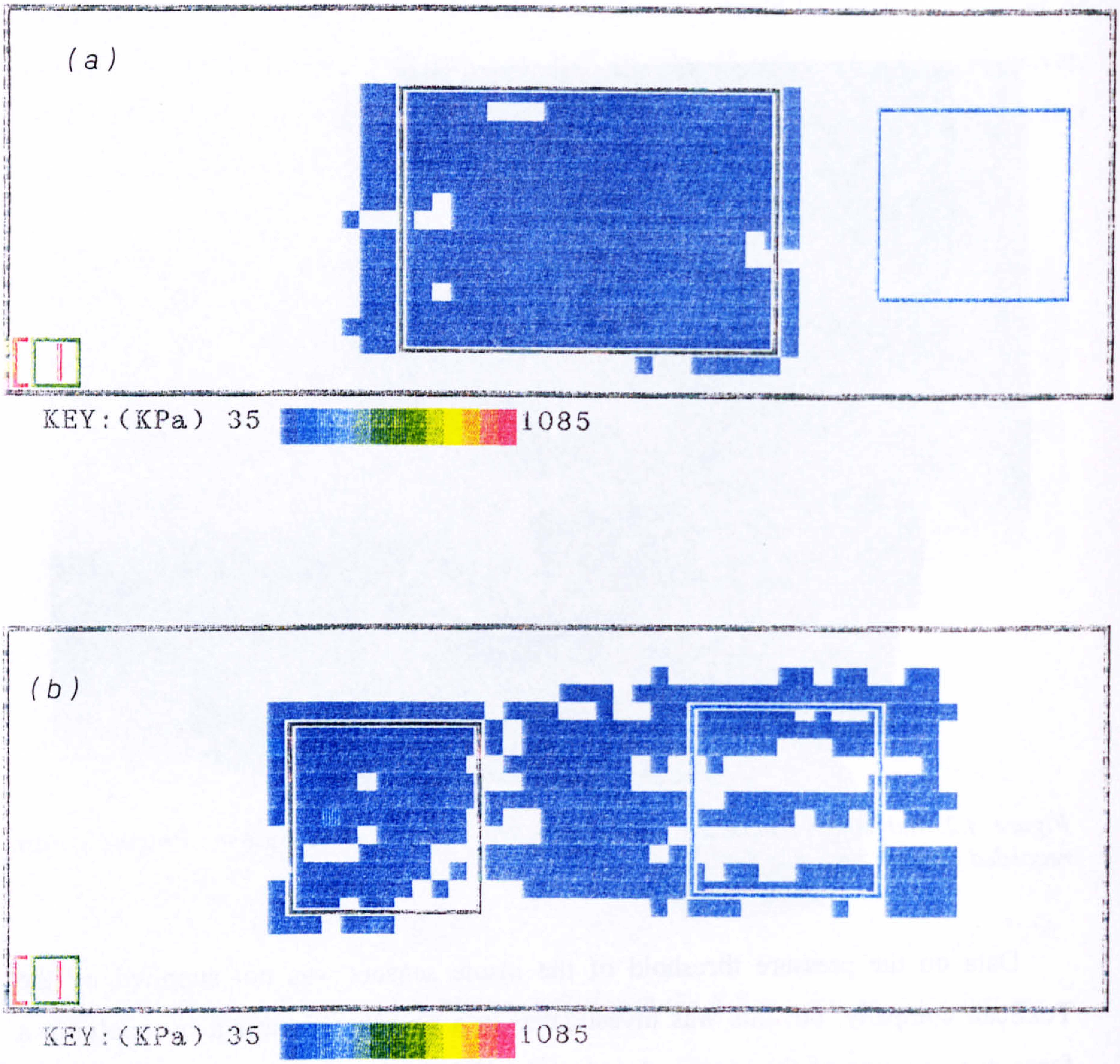

Figure 4.3. Determining the pressure threshold of the F-Scan sensor: (a) 30 kg mass applied over an area of 0.01 m^2, ie. pressure of 30 kPa (b) 30 kg mass applied over an area of 0.0152 m^2, ie. pressure of 20 kPa."

After a pressure recording has been made, the F-Scan software allows the data to be analysed in a variety of ways:

Playback mode. This mode displays the pressure recording in a two-dimensional (2-D) form (see figure 5.3). The recording can be played through continuously as a movie or a frame at a time. Colour changes on the display as the recording is played indicate pressure changes, with each colour representing a pressure range. It is optional whether a force-time graph is displayed below the pressure recording.

Pressure vs time mode. In this mode, a maximum of four boxes can be placed over user-selected areas of the 2-D pressure display and the pressure within each box displayed against time on a graph (see figure 2.16b). The size of the boxes can be varied from 15 mm x 15 mm (3 x 3 loadcells) to 105 mm x 105 mm (21 x 21 loadcells). The version of the F-Scan software used in this study (version 3.421) did not provide a function for the automatic calculation of *impulse*, ie. the force-time integral within each box. The impulse is thought to be more informative than the magnitude of the maximum peak pressure alone since it provides an indication of the magnitude of the load combined with the duration of weight-bearing. Although it was possible to calculate the impulse manually for the loading at each site, this was not done because of the time-consuming nature of this analysis.

Peak mode. Although termed the *peak* mode in the software, the 2-D display presented in this mode is, in the terminology of this thesis, the maximum peak pressure recorded by each loadcell during a single footstep. The recording can be played through one footstep at a time and the centre of pressure path overlaid (see figure 5.2).

3-D mode. The pressure recording can be viewed and played through in 3-D form (see figure 6.3). The magnitude of pressure recorded by a loadcell is proportional to the height of the display. Pressure measurements cannot be made from this display, but in some cases areas of high pressure are more readily apparent.

During the clinical trial, in-shoe plantar pressures were recorded beneath both feet simultaneously. The inner surface of the shoe in contact with the sole of the foot was standardised by inserting a special inlay into the stock shoes worn by the subjects and replacing the normal moulded or cushioning insoles in the patients' shoes. The inlay was made of two materials to provide a rigid surface beneath the forefoot and heel, and flexibility beneath the midfoot (figure 4.4a). A detailed description of this inlay and its fabrication is provided in the following section (4.3.3). A new F-Scan pressure sensor was used with each shoe. Individuals wore nylon hold-ups to prevent the foot from sticking to the sensor and causing it to crease when the shoe was put on and during a test.

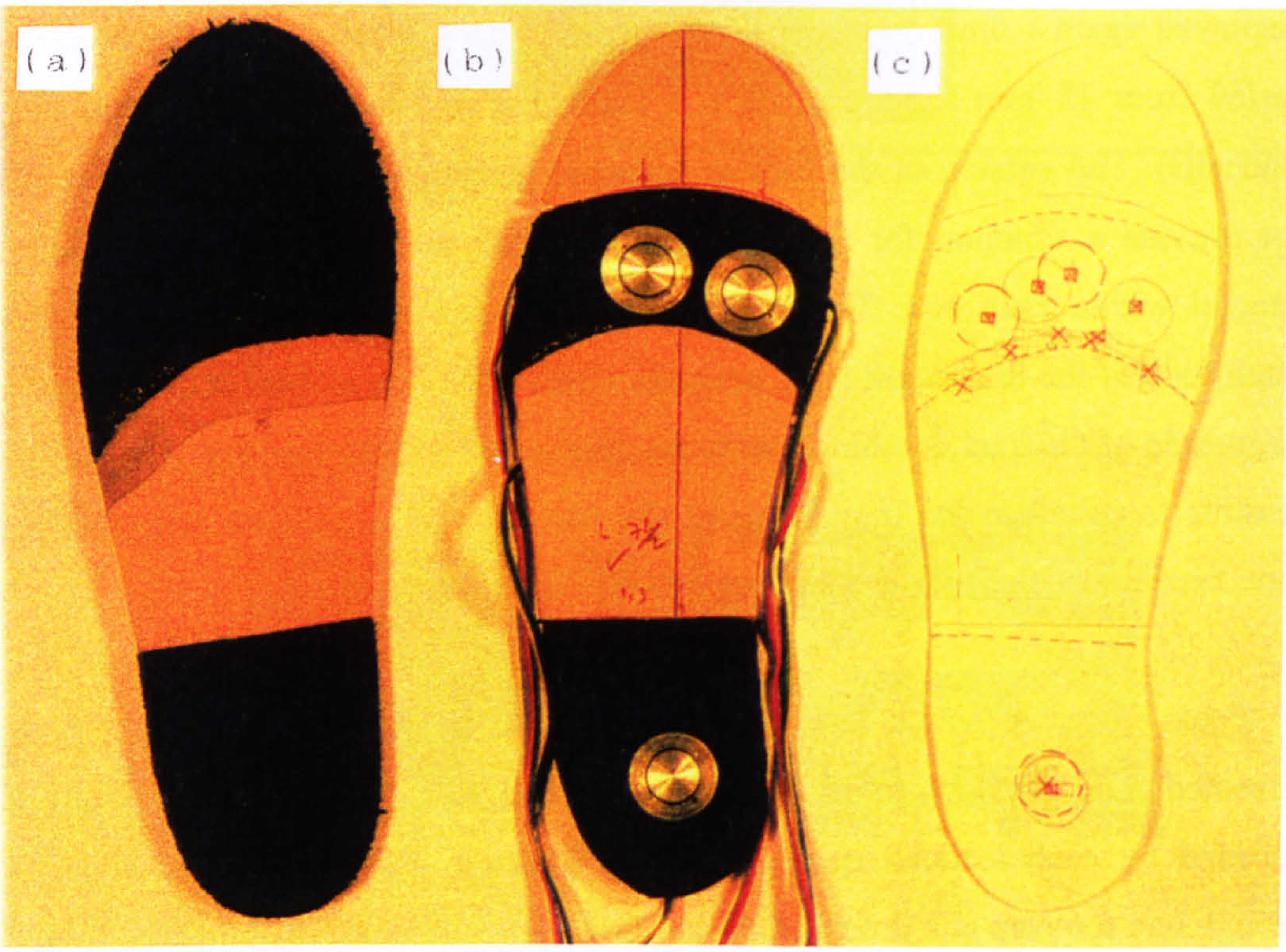

Figure 4.4. Plastazote-Poron inlays used during (a) pressure and (b) shear measurements and (c) the paper template used to guide their construction.

Individuals walked several paces in the instrumented shoes prior to data recording. The reason for doing this was threefold: firstly, to allow individuals to become accustomed to walking with the surrounding instrumentation; secondly, to allow the temperature of the sensors to equilibrate (see Chapter 6, section 6.2.2); and thirdly, to condition the sensors. TekScan Inc. recommend twelve gait cycles be completed to condition a new sensor, prior to which pressure readings will be too high. Calibration, following conditioning, was performed using the manufacturers recommended method. This required individuals to stand with all weight on the sensor in order to equate sensor output with applied bodyweight (see Chapter 6, section 6.2.2). Two 4 second pressure recordings were made to maximise the chance of a good record in which there were several successive uniform steps. Individuals walked along a corridor in a straight line in order to reduce step-to-step variation in plantar loading. Data was recorded over the central part of the walk at a rate of 100 Hz per loadcell. The walking speed was not controlled in this study as pacing by artificial means is known to result in a more erratic gait, in particular a greater variability in the contact time of successive stance phases (**Hughes et al, 1991b**). The actual walking speed of individuals was not measured as there was no accurate and repeatable method available for doing this. The *cadence*, which is proportional to the walking speed, could instead be calculated from the data contained in the pressure recordings to determine the constancy of walking between individuals.

4.3.3 shear stress measurement

Shear stresses were recorded using the transducer described in the previous Chapter. Measurements were made at the sites of peak pressure beneath the medial four metatarsal heads and heel.

The shear transducers were mounted flush into an inlay to minimise perturbation. The inlay was a four-part construction of rigid high-density Plastazote® beneath the ball area and heel, and flexible Poron® beneath the midfoot and toes (figure 4.4b). The transducer was mounted in an incompressible material so that it would remain flush with its surroundings during loading. The thickness of the inlay was determined by the nominal allowance of the stock shoes (5.4 mm) and the commercially available thicknesses of Plastazote. Plastazote 8.2 mm thick was used to produce an inlay which was comfortably accommodated in the stock shoes and was easily accommodated in the patients shoes as a replacement to their normal moulded or cushioning inserts. Care was taken not to extend the band of Plastazote in the forefoot of the inlay posterior to the metatarsophalangeal break line. The position of the join between the Plastazote and

Poron in the region of the forefoot was referenced to ink marks on a paper insole, which were transferred from the palpated plantar aspects of the metatarsal heads as an individual walked in their test shoes (figure 4.4c). The scarf joint between the materials was directly under these marks. The surfaces of the materials to be joined were ground at an angle and glued together using a plastics/polyolefin glue (3M Spray 90 high strength adhesive[30]); excess Poron on the top of the inlay was ground flat (figure 4.5). The Plastazote in the heel region made-up approximately ⅓ of the overall length of the inlay and was glued to the Poron in the midfoot using a butt joint. A butt joint was also used between the Poron and Plastazote beneath the proximal phalanx of the toes.

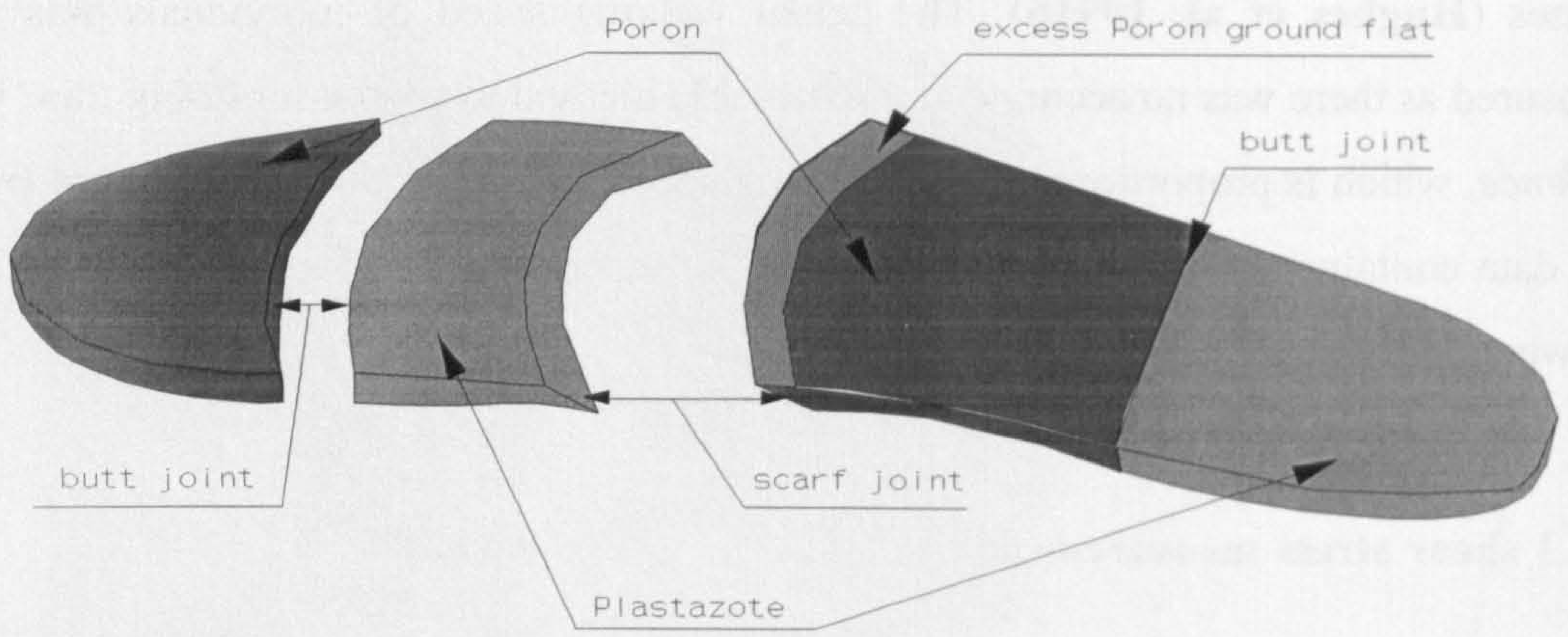

Figure 4.5. Construction of the Plastazote/Poron inlay used during pressure and shear measurements.

A housing was required to accommodate and locate the shear transducer in the inlay such that:

- the bottom stage of the transducer was fixed relative to the inlay;
- the transducer could be easily removed from one inlay for use in another;
- there was a clearance all around the transducer to allow for a top stage movement of 0.5 mm corresponding to a shear force of up to ±40 N;
- the signal leads entering the transducer were kept horizontal to prevent excessive bending that may cause them to break;
- the housing did not reduce the resilience of the inlay to compression, which

[30] 3M UK Plc, 3M House, Bracknell, Berkshire, RG12 1JU.

would subsequently allow the housing to come loose;

and

- the housing was rigid so that it did not bend during flexion of the inlay and hinder the shearing action of the transducer.

An engineering drawing of the housing that was designed is shown in figure 4.6. The two-part assembly of *collar* and *baseplate* were machined from an aluminium alloy. A tough instant adhesive (*Loctite 405*[31]) was used to glue the bottom stage of the transducer to the baseplate and a disc was glued to the top stage of the transducer to bring it flush with the surrounding Plastazote (figure 4.7). The collar and baseplate were held together with screws and the whole assembly was securely located in the inlay by trapping an annulus of Plastazote. The positioning of the transducers in the inlay and shoe is shown in figures 4.4b and 4.8, respectively.

[31] Loctite UK, Welwyn Garden City, Hertfordshire AL7 1JB, UK.

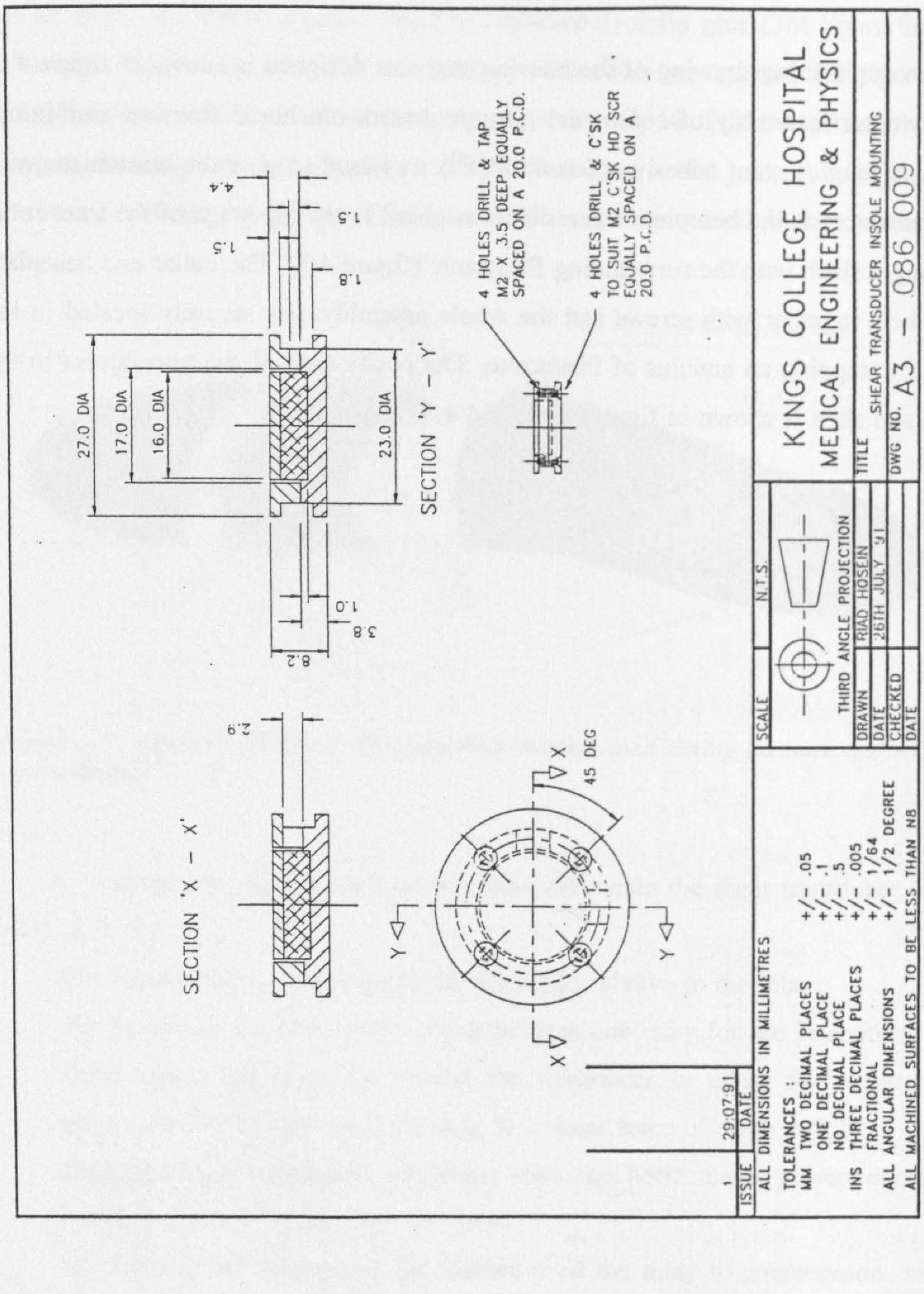

Figure 4.6. Engineering drawing of the housing designed to accommodate the shear transducer in the Plastazote-Poron inlay.

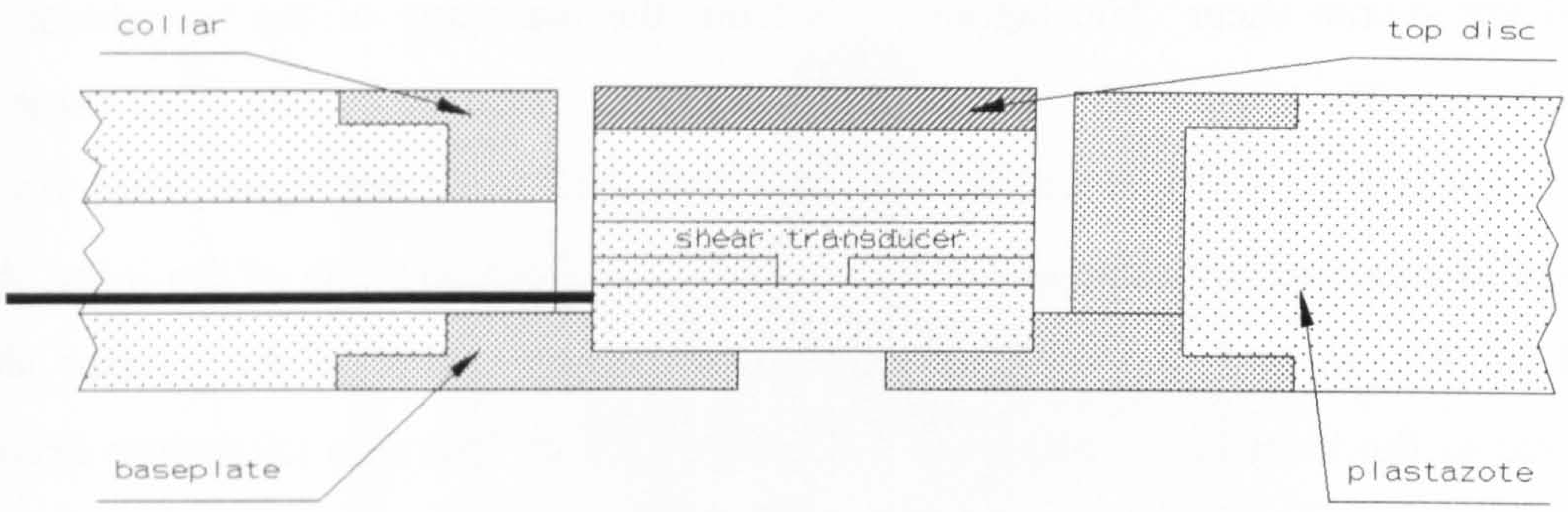

Figure 4.7. Schematic of the shear transducer mounted in the Plastazote-Poron inlay.

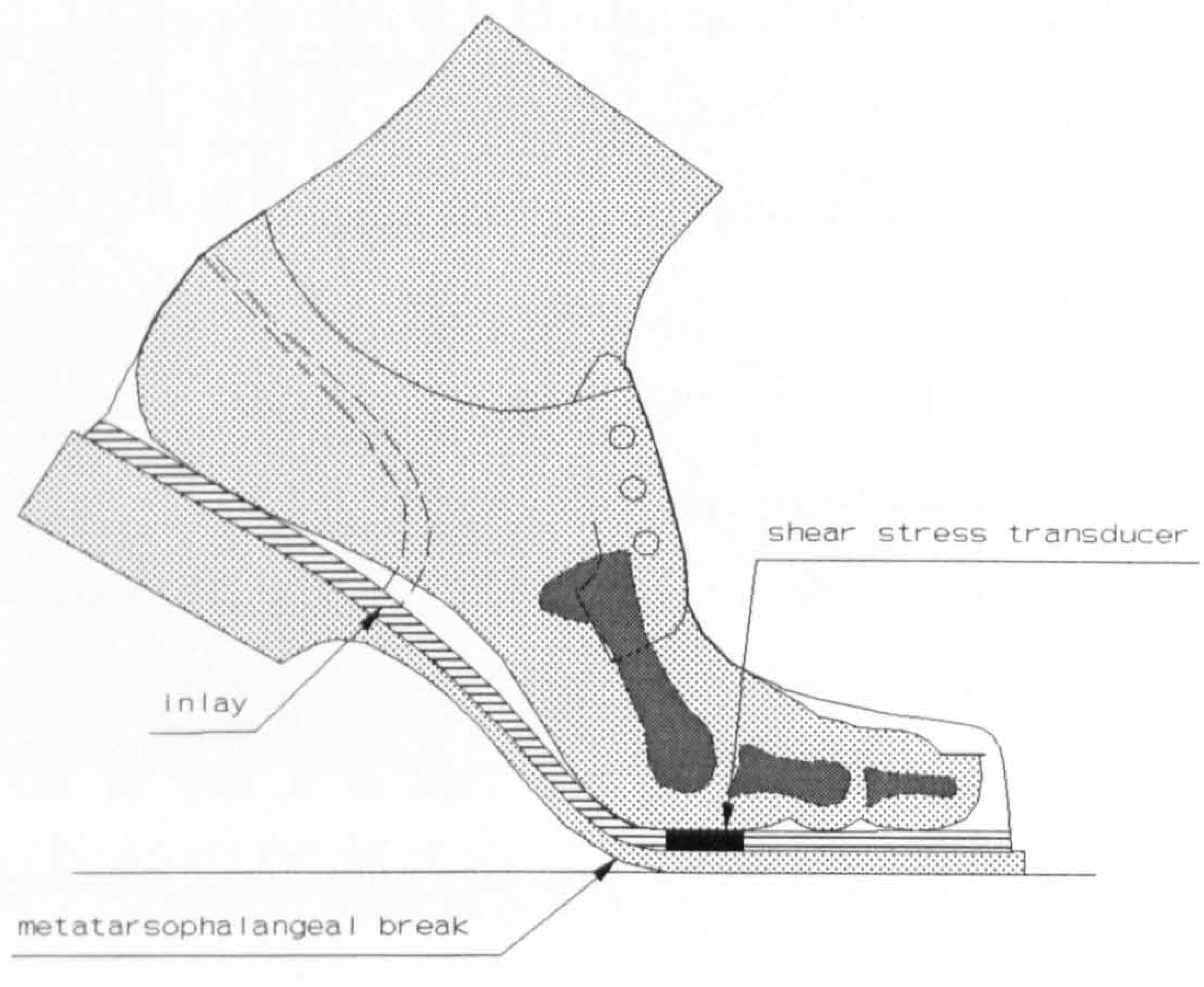

Figure 4.8. Location of the shear stress transducer in the forepart of the shoe ahead of the metatarsophalangeal break.

Because of the close proximity of the areas of high pressure associated with the medial four metatarsal heads, four shear transducers could not be placed side-by-side in this region. Instead two inlays were made for each shoe: in one inlay, transducers were placed at the sites of peak pressure beneath the first and third metatarsal heads and heel; and in the other inlay, transducers were placed at the sites of peak pressure beneath the second and fourth metatarsal heads and heel. Two walks were necessary to record plantar shear stresses beneath the medial four metatarsal heads. The duplicate measurements made beneath the heel were useful in assessing the consistency of gait and the consistency of the shear stresses generated after a change of inlay. The sites of maximum peak pressure beneath the medial four metatarsal heads and heel have been marked on the paper insole shown in figure 4.4c (crosses) and the placement of the shear transducers outlined (circles).

The top stage of the transducer was orientated to record longitudinal shear and was aligned with the midline of the foot. This left the bottom stage of the transducer to record transverse shear. The signal leads from the top stage of the transducer left through the sides of the inlay, whereas the signal leads from the bottom stage came out underneath (figure 4.4b). It can be seen in figure 4.4(b) that the signal leads for the longitudinal axis of each transducer did not all leave from one side of the inlay. As a result of this, in two of the three transducers the application of an anterior shear (relative to the transducer) produced a negative output. This was taken into account when the output from the longitudinal axis of each transducer was plotted by adopting a sign convention for the application of an anterior shear to the transducer.

A *standard* shear test constituted four recorded walks with each instrumented inlay: two with the individual not wearing hosiery and two wearing nylon hold-ups (15 denier). Duplicate recordings were made to maximise the chance of a good record in which there were several successive uniform steps. The hold-ups were chosen to alter the surface friction conditions without taking up significant volume in the shoe. Measurements were made three months apart on one subject and one patient to estimate differences over time. For each test the individual walked in a straight line along the length of a 10 m laboratory (figure 4.9). Data was recorded for 5 seconds over the central part of the walk at a rate of 400 Hz per transducer axis and footswitch. Similar to the pressure recordings, the walking speed was not controlled.

Figure 4.9. Subject instrumented with the system for in-shoe plantar shear stress measurement. The backpack contains the transducer signal-conditioning electronics.

4.3.4 biomechanical examination

Foot mobility and fixed foot deformities are thought to influence the plantar stress distribution, but the extent to which this is the case is not known. A biomechanical assessment was conducted to measure joint ranges of motion and assess the nature of any fixed foot deformities, so that the relationship between these measures and the patterns and magnitudes of in-shoe plantar stresses could be determined.

The procedure for the biomechanical examination is outlined in Chapter 2, section 2.2.3. The chart on which measurements were recorded is shown in figure 4.10. Intra-observer measurements were made on two subjects on four consecutive days to assess repeatability.

Name:	Patient/subject:		Date:	
			Right foot	Left foot
lying (knee extended)	ankle joint	neutral forced dorsiflexion	___ ___ ___	___ ___ ___
	subtalar joint	neutral inversion eversion	___ ___ ___	___ ___ ___
	midtarsal joint	forefoot varus forefoot valgus	___ ___	___ ___
	1st metatarsal	dorsiflexed plantarflexed	___ ___	___ ___
	5th metatarsal	dorsiflexed plantarflexed	___ ___	___ ___
standing		tibial varum tibial valgum	___ ___	___ ___
		rearfoot varus rearfoot valgus	___ ___	___ ___
		angle of gait		

Figure 4.10. Biomechanical examination record chart.

4.3.5 neurological assessment of patients with diabetes

Vibration perception was quantitated to assess the response of the large myelinated fibres of the foot and hence the degree of sensory neuropathy. A *Bio-thesiometer*[32] was used for this assessment (figure 4.11). This is a hand-held instrument that has a plastic protrusion which can be controlled to vibrate with varying amplitude. The amplitude of vibration is recorded in arbitrary units of *volts*.

With the patient sitting-up on an examination couch, the instrument was first operated on one of the patient's hands where there was some sensory perception, in order to indicate the vibrating sensation. Measurements were then made at two sites on each foot, first with the vibrator placed on the apex of the hallux and then on the

32 Bio-medical Instrument Company, 15764 Munn Road, Newbury, Ohio 44065, USA.

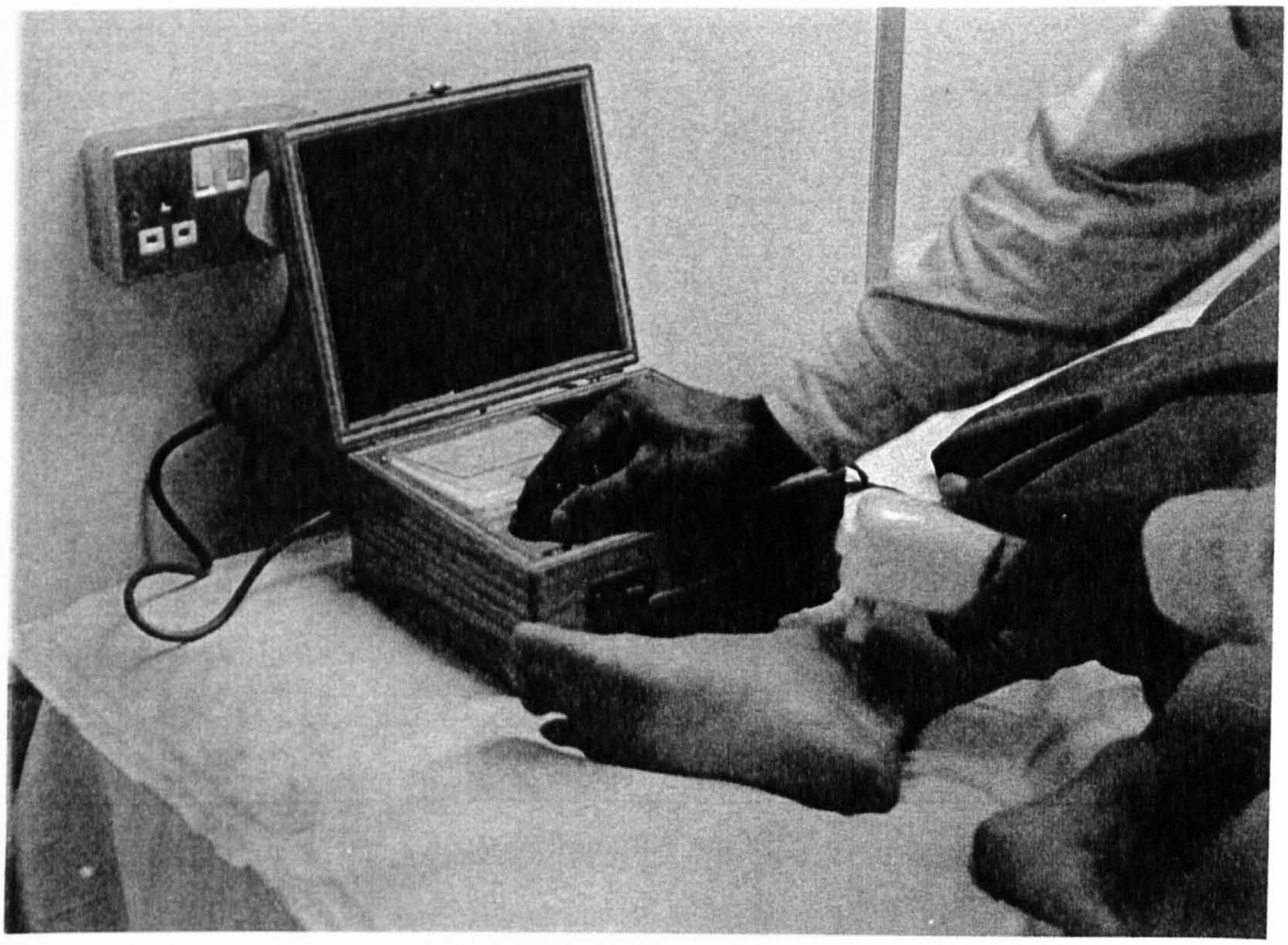

Figure 4.11. Method of assessing the degree of peripheral sensation in the feet. The assessment is being carried out on the medial malleoli of the right foot.

medial malleolus (figure 4.11). The weight of the instrument provided the contact force. The amplitude of vibration was increased by increasing the voltage supply and the patient indicated as soon as their vibration threshold was reached. Three measurements were made at each site to obtain an average. A threshold of less than 11 volts at the apex of the hallux (**Foster and Edmonds, 1987b**) and the medial malleolus is sufficient to provide normal protective sensation. **Foster** states that the vibration threshold should be similar at both sites (personal communication). However, as discussed in Chapter 2, section 2.6.2, sensory neuropathy begins distally and progresses proximally, so it is reasonable to expect greater sensory impairment in the more distal regions of the foot in some patients.

CHAPTER 5
RESULTS

5.1 INTRODUCTION

This Chapter presents the characteristics of the study groups and the results of the clinical trial described in Chapter 4.

Univariate statistical analyses were performed on the data: correlations and statistical significance were assessed using the *Pearson Product Moment* and *t-test*, respectively. The assumption of normality required by the t-test was checked using the *Kolmogorov-Smirnov (K-S) test*, which established all subject and patient data *was* normally distributed (ie. $p>0.05$) (see Appendix A). In tests of statistical significance, differences at less than the 5% level were considered to be significant, ie. $p<0.05$.

Contralateral feet were not treated as independent in tests of statistical significance - contralateral feet are highly related statistically (**Wenger et al, 1989** and letter to the editor). For this reason, any data that was gathered from the right and left foot of an individual was first averaged before use in tests of statistical significance (**Rosner, 1982**).

5.2 SUBJECT AND PATIENT CHARACTERISTICS

A total of nine asymptomatic subjects (eight male) and seven diabetic patients (five male) were recruited to the trial. In the subject group, ages ranged from 25 to 42 years (mean 34.3, sd 6.5 years) and body mass from 65.1 to 90.8 kg (mean 77.1, sd 10.1 kg); while in the patient group, ages ranged from 48 to 78 years (mean 63.6, sd 11.8 years) and body mass from 78 to 108 kg (mean 90.6, sd 12.4 kg). The characteristics of individual subjects and patients appear in tables 5.1 and 5.2, respectively. The patients were significantly older ($p=0.0003$) and heavier ($p=0.039$) than the subjects (unpaired t-test).

Table 5.1. Characteristics of the subject group.

Subject	Age (years)	Body mass (kg)	Sex	
1	41	65.1	m	
2	28	67.6	m	
3	25	67.2	m	
4	40	72.0	m	
5	39	90.8	m	
6	28	82.0	m	
7	31	90.0	m	
8	42	74.0	f	
9	35	85.0	m	

Table 5.2. Characteristics of the patient group.

Patient	Age (years)	Body mass (kg)	Sex	Type of diabetes (niddm/iddm)
1	48	105.0	m	niddm
2	73	108.8	m	iddm
3	53	89.2	f	iddm
4	75	81.5	m	niddm
5	56	78.1	f	niddm
6	78	78.0	m	iddm
7	62	94.3	m	iddm

The subjects had no history of lower limb pathologies. On the plantar surface of the feet there was a normal light covering of callus around the border of the heel and on the medial border of the first metatarsophalangeal joint. In subject number 8 there were areas of thick plantar callus between the first and second and the second and third metatarsal heads and over the fourth.

The seven patients were recruited to the trial over a seven month period. Verbal consent for participation was obtained in each case. Considerably more difficulty was encountered in recruiting the patients because of the strict selection criteria (see Chapter 4, section 4.3.1). A total of two-hundred-and-five patients were seen in the

Diabetic Foot Clinic at King's College Hospital during the seven month period. Patients attending the London Foot Hospital[33] were also assessed for inclusion, but out of the two-hundred-and-twenty that were seen there, only one met the acceptance criterion. Unfortunately this patient was not willing to take part in the study. Most of the patients that were seen at both hospitals were excluded because they had no history of plantar metatarsal ulceration. Of those that did, a considerable number were excluded because they also had digital amputations of the same foot, which may have resulted in altered foot function compared to that of an intact foot. A total of ten healed ulcers were noted on the plantar surface beneath the forefoot of the seven patients. The most common site of ulceration was beneath the first metatarsal head, followed by the second (table 5.3). One patient also had seed corns. These are hard plugs of cholesterol with no surrounding callus formation that only occur on the sole of the foot regardless of high pressure, but usually in the presence of anhidrosis (deficient sweat secretion). The right foot of patient number 3 was not included in any of the tests because the second, third, fourth and fifth metatarsal heads had been surgically removed. All patients had been supplied with bespoke shoes: one had rocker-bottom shoes and all except one had rocker-inserts (table 5.3). Patient occupations were also noted and are presented in table 5.3. Common to all occupations was a necessity for standing or walking for long periods throughout the day.

[33] London Foot Hospital, 33 Fitzroy Square, London W1P 6AY, UK.

Table 5.3. Plantar problems and shoe characteristics of the patients.

Patient	Most recent occupation	Sites of plantar ulceration		Sites and nature of other plantar problems		Clawed/retracted hallux	Clawed/retracted lesser toes	Rocker-bottom shoes	Rocker inserts
		left foot	right foot	left foot	right foot				
1	train driver	1st MTH	1st MTH	-	-	x	x	x	✓
2	garage owner (now retired)	-	2nd MTH	-	-	x	x	x	✓
3	housewife	1st MTH; heel: posterior aspect	not tested	-	-	✓	✓	x	✓
4	optical mechanic (now retired)	1st and 5th MTHs	-	-	heavy callus 1st MTH	x	x	✓	✓
5	cleaner	2nd/3rd MTH area	-	-	-	✓	✓	x	✓
6	tea merchant: importer (now retired)	1st MTH	-	-	seed corns 3rd and 4th MTHs	✓	✓	x	x
7	civil servant: post room worker	1st MTH	2nd MTH	-	callus 1st MTH	x	x	x	✓

All except one patient had peripheral sensory neuropathy in both feet. Vibration thresholds measured during the neurological assessment are presented in table 5.4. The figures quoted are in *arbitrary* units of volts as presented by the *Bio-thesiometer* and are the averages of three tests carried out at each site. Recall that a threshold of less than 11 volts is sufficient to provide normal protective sensation, as discussed in Chapter 4, section 4.3.5. The patients had varying degrees of neuropathy. Variation was also noted between contralateral feet and within feet at the two sites where measurements were made.

Table 5.4. Average vibration thresholds (± 1sd, volts) in the feet of the patients, assessed using the Bio-thesiometer.

Patient	Left foot		Right foot	
	medial malleoli	hallux	hallux	medial malleoli
1	21.5 (3.0)	9.2 (5.5)	13.7 (5.7)	13.3 (3.8)
2	off scale (>50)	35.7 (6.5)	37.0 (8.9)	off scale (>50)
3	off scale (>50)	off scale (>50)	off scale (>50)	off scale (>50)
4	off scale (>50)	20.3 (4.9)	13.7 (4.0)	22.7 (4.0)
5	19.5 (1.8)	18.0 (1.0)	9.3 (3.1)	32.0 (1.0)
6	15.3 (2.5)	15.0 (0.9)	29.2 (5.6)	22.3 (1.2)
7	3.3 (0.6)	5.7 (1.5)	7.3 (1.5)	4.7 (0.6)

5.3 PRESSURE MEASUREMENT

As mentioned in Chapter 4, pressure recordings were made to locate the sites of peak pressure beneath the metatarsal heads and heel so that the shear transducers could be placed in these regions of maximum load. It was decided from the outset of the analysis to identify these pressure peaks from the *second* complete gait cycle in the *second* pressure recording that was made with each foot. In a few cases, this particular recording could not be analysed because the F-Scan sensor had become creased during walking, caused by sliding against the Plastazote of the inlay inside the shoe, which resulted in whole lines of loadcells to stop registering pressure. Where this was the case, the *first* pressure recording was analysed instead. In these individuals, the pressure data analysed had not been obtained from beneath both feet simultaneously.

As a result, the pressure data from these subjects and patients was not included in any tests of statistical significance.

Several other analyses were performed on the data collected by the F-Scan system:

- the *cadence*, ie. the number of gait cycles (steps) per minute, was calculated to determine the constancy of walking between individuals of the two groups;
- the percentage durations of the three temporal phases of gait were calculated to determine whether there were any deviations from the values considered normal;
- the locus of the centre of pressure (CoP) was analysed to determine if there were any deviations from the path considered normal;

and

- the values and timings of maximum peak pressure in the region of the hallux, medial four metatarsal heads and heel were measured to determine whether there were any differences between the two groups and differences between other in-shoe and barefoot studies.

5.3.1 cadence

Cadences were calculated for each individual based on the duration of each complete step (see Appendix B). In most cases pressure was recorded from beneath each foot over six complete steps; while pressure was recorded over fewer steps in slower walkers.

The figures in Appendix B were used to calculate an average cadence for each individual. These averages, which are presented in table 5.5, were used in statistical analyses. Subject and patient cadences were similar (p=0.15, unpaired t-test)[34], ie. the subjects and patients walked at the same rate while in-shoe plantar pressures were being recorded. Even so, there was a greater variation amongst the patients compared to the subjects as indicated by the larger standard deviation of the patients' group mean.

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[34] Data from subjects 2, 8 and 9, and patients 3 and 7 was not included in this significance test because plantar pressure recordings were not obtained from both feet simultaneously.

Table 5.5. Average cadences of the subjects and patients during plantar pressure measurements (steps/minute (± 1sd)).

Subject	Average cadence (sd)	Patient	Average cadence (sd)
1	113.1 (3.2)	1	110.1 (4.6)
2[a]	113.1 (2.0)	2	94.2 (0.8)
3	112.2 (1.7)	3[b,c]	106.5 (6.4)
4[b]	106.6 (1.7)	4	117.5 (1.7)
5	112.3 (0.8)	5	102.4 (4.3)
6	117.2 (2.1)	6[b]	77.8 (2.0)
7[b]	115.0 (1.5)	7[a]	94.1 (6.7)
8[a]	107.8 (2.7)	-	-
9[a]	97.8 (2.6)	-	-
mean (sd)	112.7 (3.6)[d]	mean (sd)	100.4 (15.3)[e]

[a] pressure recordings were not obtained from both feet simultaneously.
[b] analysis performed on the first set of in-shoe pressure recordings which were made.
[c] pressure recordings were not obtained from beneath the right foot.
[d] mean of data from subjects 1, 3, 4, 5, 6 and 7, from whom pressure recordings were obtained from both feet simultaneously.
[e] mean of data from patients 1, 2, 4, 5 and 6, from whom pressure recordings were obtained from both feet simultaneously.

5.3.2 timing of gait phases

The three temporal phases of gait, *stance*, *swing* and *double-support*, are considered to last for 62%, 38% and 12% of a gait cycle, respectively (**Mann, 1991**) (see figure 2.14). The duration of the stance and swing phases was calculated for each complete gait cycle from the pressure records; the duration of the double-support phase was only calculated for the subjects and patients from whom pressure recordings had been obtained from both feet simultaneously. The results, which are presented in Appendix C, reflect step-to-step variation in the duration of gait cycle events for one foot, as well as between contralateral feet. An average duration of each phase was calculated for each person by averaging the right and left foot data presented in Appendix C. These results, which are presented in tables 5.6 and 5.7 for the subjects and patients, respectively, were used in statistical analyses.

Table 5.6. Average durations of the stance, swing and double support phases during plantar pressure measurements, expressed as a percentage of the gait cycle (± 1sd), in the subject group.

Subject	Stance	Swing	Double support
1	63.8 (1.8)	36.2 (1.8)	13.7 (2.0)
2[a]	60.6 (1.3)	39.4 (1.3)	-
3	60.6 (1.8)	39.4 (1.8)	10.7 (0.6)
4[b]	63.9 (1.0)	36.1 (1.0)	13.0 (0.4)
5	63.0 (0.8)	37.0 (0.8)	12.9 (1.6)
6	62.5 (0.8)	37.5 (0.8)	12.5 (0.9)
7[b]	61.4 (1.4)	38.6 (1.4)	11.2 (1.2)
8[a]	62.8 (1.7)	37.2 (1.7)	-
9[a]	63.3 (1.0)	36.7 (1.0)	-
mean (sd)[d]	62.5 (1.3)	37.5 (1.3)	12.3 (1.2)

Table 5.7. Average durations of the stance, swing and double support phases during plantar pressure measurements, expressed as a percentage of the gait cycle (± 1sd), in the patient group.

Patient	Stance	Swing	Double support
1	62.5 (1.4)	37.5 (1.4)	12.5 (1.2)
2	69.3 (3.4)	30.7 (3.4)	19.0 (2.7)
3[b,c]	63.1 (2.2)	36.9 (2.2)	-
4	66.5 (2.8)	33.5 (2.8)	17.2 (2.7)
5	64.5 (1.0)	34.9 (0.8)	14.0 (1.3)
6	64.1 (1.3)	35.9 (1.3)	14.1 (1.7)
7[a]	65.5 (1.1)	34.5 (1.1)	-
mean (sd)[e]	65.4 ± 2.6	34.5 ± 2.6	15.4 ± 2.7

[a] pressure recordings were not obtained from both feet simultaneously, as a result the duration of the double-support phase could not be calculated.
[b] analysis performed on the first set of in-shoe pressure recordings which were made.
[c] pressure recordings were not obtained from beneath the right foot.
[d] mean of data from subjects 1, 3, 4, 5, 6 and 7, from whom pressure recordings were obtained from both feet simultaneously.
[e] mean of data from patients 1, 2, 4, 5 and 6, from whom pressure recordings were obtained from both feet simultaneously.

The durations of the stance, swing and double-support phases were similar between the two groups ($p=0.078$, 0.067 and $p=0.064$, respectively, unpaired t-tests)[34]. However, based on the single average calculated for the duration of each phase it is apparent that the values for the patient group are not comparable with those considered normal (see tables 5.7) - the stance and double-support phases recorded were longer. Also apparent is the larger variation in the duration of each phase between individual patients, as indicated by the larger standard deviations.

The walking speed of elderly men (aged 67 to 87) has been found to be significantly lower than the walking speed of younger men (aged 20 to 65) (**Murray et al, 1969**). Based on this finding it is reasonable to suppose that walking may become gradually more tentative and hence gradually slower with ageing. In order to test this hypothesis age was correlated with the duration of the stance phase in both groups. A positive correlation between age and the duration of the stance phase was found in both groups, but this was not significant in the patient group (subjects: $r=0.835$, $0.001<p<0.01$; patients: $r=0.652$, NS) (figure 5.1).

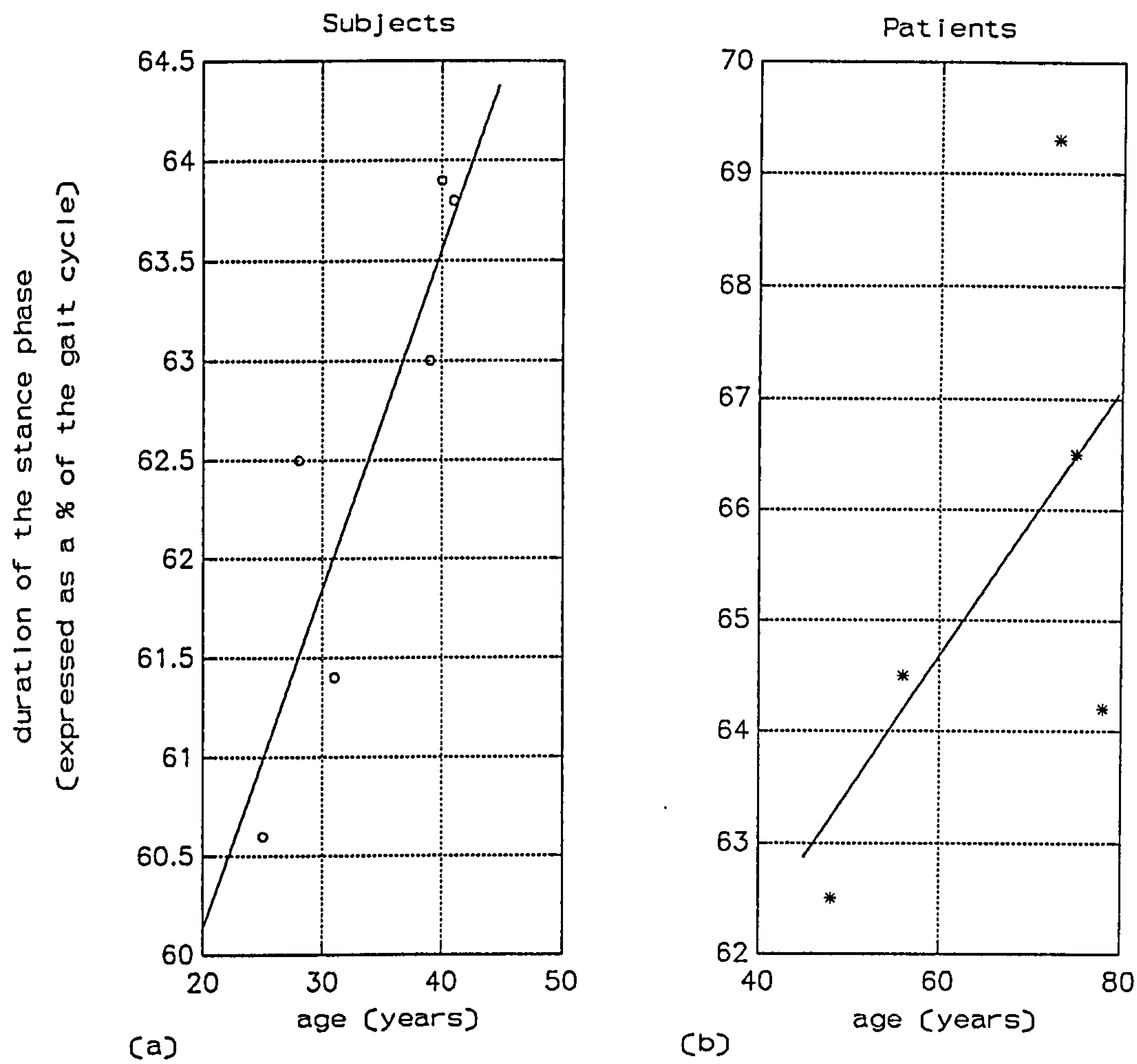

Figure 5.1. Correlation between age and the duration of the stance phase in the subject and patient groups.

5.3.3 centre of pressure

Tables 5.8 and 5.9 provide descriptions of the locus of the CoP for the second complete step taken by the subjects and patients, respectively, during in-shoe plantar pressure measurements. Successive numbers have been used to indicate the progression of the locus with respect to the anatomical structures of the foot. Descriptions are not included in the few cases where creasing of the F-Scan sensor inside the shoe caused loadcells to stop registering pressure, which resulted in erroneous plots of the locus. With the exceptions of subjects number 2, 8 and 9 and patients number 3 and 7, plantar pressure, and hence loci, were obtained from both feet simultaneously. Two plots are presented in figure 5.2 showing the locus of the CoP in-shoe for one subject and one patient. The boxes in these plots bound areas of peak pressure associated with the first, second, third and fourth metatarsal heads.

Table 5.8. Path of the locus of the centre of pressure beneath the feet of the subjects.

Subject /foot	Origin				Midfoot/forefoot travel						End			
	posterior heel	postero-lateral heel	heel centre	anterior heel	medial to midline	along midline	lateral to midline	beneath mth4	beneath mth3	beneath mth2	mth2	anterior to mth2	proximal aspect of hallux	lateral side of hallux
typical path		1					2	3	4	5				6
1-L			1				2			3			4	
1-R			1			2					3			
2-L			1				2	3	4	5				6
2-R			1				2		3	4				5
3-L				1			2		3			4		
4-L			1			2				3				4
4-R			1			2								3
5-R			1			2					3			
6-L			1				2		3			4		
6-R			1				2				3			
7-L				1			2			3				4
7-R			1			2			3	4			5	
8-L			1			2				3			4	
8-R			1				2						3	
9-L			1				2			3		4		
9-R			1				2		3		4			

Table 5.9. Path of the locus of the centre of pressure beneath the feet of the patients.

Patient /foot	Origin				Travel						End		
	posterior heel	postero-lateral heel	heel centre	anterior heel	medial to midline	along midline	lateral to midline	beneath mth4	beneath mth3	beneath mth2	mth2	anterior to mth2	lateral side of hallux
typical path		1					2	3	4	5			6
1-L			1				2			3			4
2-L	1						2			3		4	
2-R	1						2		3	4			5
3-L			1			2				3		4	
4-L			1		2						3		
4-R			1		2						3		
5-L			1			2				3		4	
5-R			1			2				3		4	
7-L			1				2		3			4	
7-R	1						2		3	4		5	

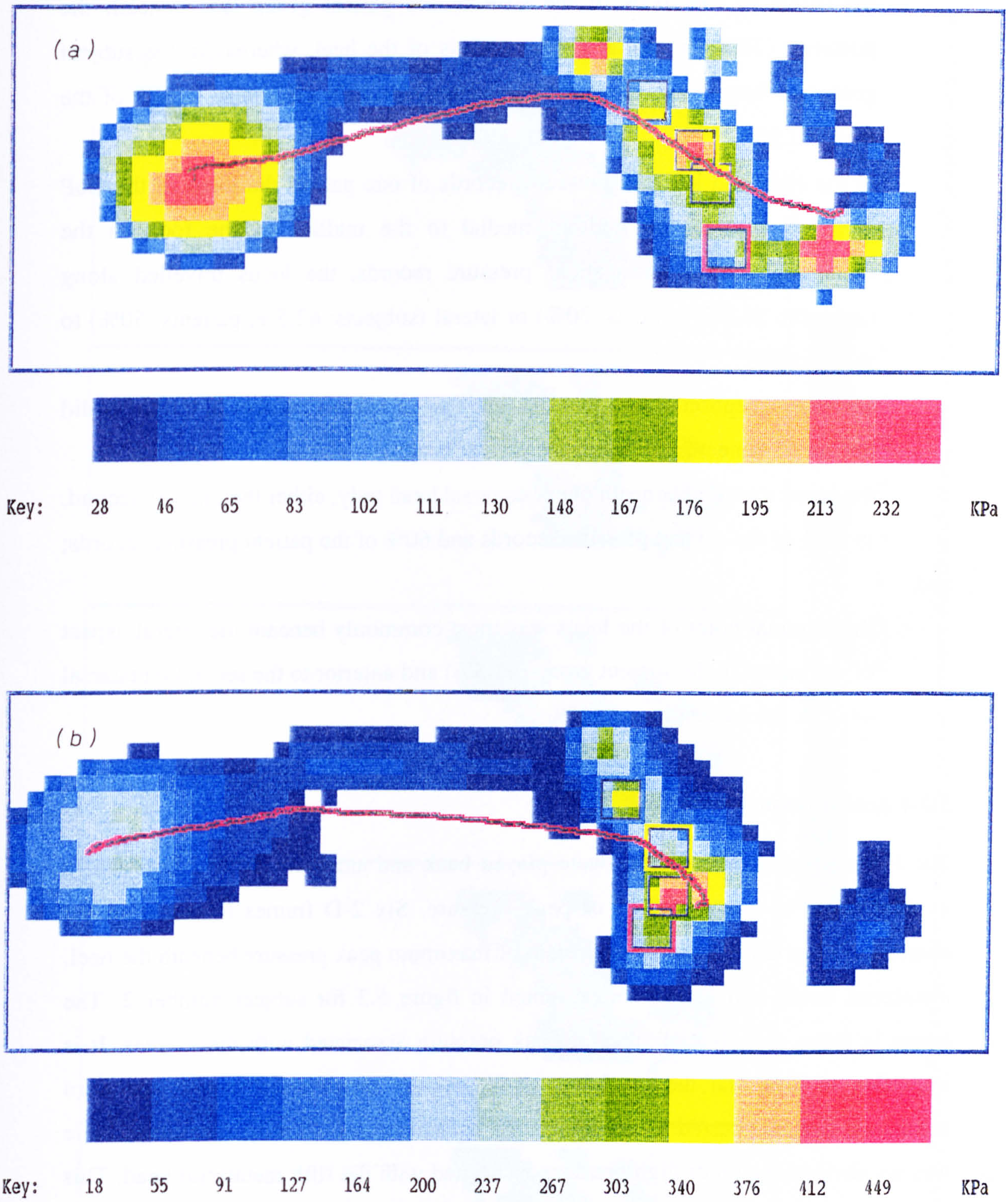

Figure 5.2. Example plots of the locus of the centre of pressure in-shoe for (a) one subject (number 2) and (b) one patient (number 7).

For each individual, the locus of the CoP in-shoe did not exactly match the typical path of the locus for barefoot walking:

- the origin of the locus of the CoP in the patient group was beneath the posterior (30%) or central (70%) aspects of the heel, whereas in the subject group the locus most commonly originated beneath the central aspect of the heel (87.5%);
- in the right and left foot pressure records of one patient the locus of the CoP travelled beneath the midfoot medial to the midline of the foot; in the remaining subject and patient pressure records, the locus travelled along (subjects: 37.5%; patients: 30%) or lateral (subjects: 62.5%; patients: 50%) to the midline;
- with the exception of one foot (the left foot of subject number 2) the locus did not travel beneath the fourth metatarsal head;
- the locus travelled beneath one metatarsal head only, either the third or second, in 50% of the subject pressure records and 60% of the patient pressure records;

and

- the terminal point of the locus was most commonly beneath the lateral aspect of the hallux in the subject group (31.3%) and anterior to the second metatarsal head in the patient group (60%).

5.3.4 peak pressure analysis

The F-Scan pressure recordings were played back and analysed in the 2-D *pressure vs time* mode to locate regions of peak pressure. Six 2-D frames representing the plantar pressure distribution at the instant of maximum peak pressure beneath the heel, metatarsal heads and hallux are presented in figure 5.3 for subject number 2. The boxes in these plots bound areas of peak pressure associated with these areas. It is important to note that the maximum peak pressure beneath the second and third metatarsal heads occurred at the same time in this subject. In most recordings there was no obvious region of high pressure associated with the fifth metatarsal head. This was a more common finding in the subject group who wore stock orthopaedic shoes. The upper of the stock shoes was wider than the sole in the forefoot, as a result the fifth metatarsal head tended to rest at the extreme edge of the pressure sensor. To be consistent, the fifth metatarsal head is not included in the following analyses.

Figure 5.3. 2-D frames from an F-Scan recording, showing the plantar pressure distribution at the instant of maximum peak pressure beneath the heel, metatarsal heads and hallux.

Local maximum peak pressures were measured over an area of 2.25 cm^2 using 15 mm x 15 mm analysis boxes (figure 5.3). This size of box was chosen as the closest to the area of the shear transducer (2 cm^2) and to be comparable in size to the discrete pressure transducers used by other workers. A pressure-time graph is automatically displayed by the F-Scan software when the analysis boxes are placed over user-selected areas of the 2-D pressure display, an example of this is shown in figure 2.16b in Chapter 2. From such graphs, the times of onset and cessation of pressure beneath each of the regions of interest were determined, along with the time at which the maximum peak pressure occurred.

The magnitudes of maximum peak pressure beneath the hallux, medial four metatarsal heads and heel, for each complete stance phase in the subject and patient pressure records, are presented in Appendix D. For each individual an average maximum peak pressure has been calculated for each site based on the data gathered from each complete stance phase in the pressure records. This data is presented in tables 5.10 and 5.11, respectively. In the subject group, the second and third metatarsal heads were jointly the sites of the highest maximum peak pressure beneath the forefoot (table 5.10); while in the patient group, the first metatarsal head was the site of maximum peak pressure followed by the second. The plantar pressures recorded beneath the foot of one patient were not suitable for analysis because creasing of the F-Scan sensor inside the shoe made the data unreliable. Of the remaining feet, the sites of four out of nine healed plantar ulcers were coincident with the sites of highest maximum peak pressure beneath the forefoot (table 5.11).

Groups averages of the maximum peak pressure at each site indicate a bias towards central loading of the forefoot in the subjects and medial loading of the forefoot in the patients. For statistical analysis, data from corresponding sites beneath the right and left foot of each individual were averaged (data from tables 5.10 and 5.11). Average maximum peak pressures beneath the hallux, heel and the first, second, third and fourth metatarsal heads were similar between the two groups (p=0.5, 0.38, 0.27, 0.38, 0.97 and 0.35, respectively, unpaired t-tests[35]).

The Pearson product moment (*r*) was calculated to correlate age and body mass with average local maximum peak pressures. In these calculations *average local maximum peak pressures* were calculated for each individual from the data presented in Appendix D by averaging the maximum peak pressure data from corresponding right and left foot sites. The Pearson product moments are presented in tables 5.12 and 5.13 for the subjects and patients, respectively. In the subject group, age was significantly positively correlated with maximum peak pressure measured beneath the first metatarsal head (r=0.753, $0.02<p<0.05$); while body mass was positively correlated with maximum peak pressures measured beneath the second, third and fourth metatarsal heads, although this was not statistically significant. In the patient group, age was not significantly correlated with local maximum peak pressures, although body mass was significantly positively correlated with maximum peak pressure beneath the hallux (r=0.916, $0.02<p<0.05$)[35].

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35 Data from patients 3 and 6 were not included in this analysis because plantar pressure data was only available from one foot.

Table 5.10. Average (± 1sd) maximum peak pressures (kPa) beneath the feet of the subjects during shod walking.

Subject	Left foot						Right foot					
	heel	mth4	mth3	mth2	mth1	hallux	hallux	mth1	mth2	mth3	mth4	heel
1	160.0 (4.6)	128.3 (11.4)	161.3 (27.2)	163.3 (24.0)	287.3 (58.1)	78.9 (13.4)	108.8 (23.3)	203.3 (51.2)	197.3 (16.7)	203.3 (23.6)	133.0 (17.7)	159.3 (4.7)
2[a]	214.7 (4.5)	129.3 (20.5)	188.3 (19.1)	177.7 (9.5)	137.7 (11.5)	256.0 (16.6)	288.0 (21.8)	229.7 (61.8)	146.7 (6.4)	187.3 (34.7)	155.3 (37.2)	226.0 (6.6)
3	167.0 (14.2)	154.7 (68.0)	191.0 (43.7)	170.0 (26.2)	131.3 (20.1)	160.7 (45.5)	96.6 (9.2)	129.3 (17.2)	144.3 (15.0)	143.3 (25.1)	111.2 (19.0)	154.0 (3.5)
4	204.3 (1.2)	134.0 (23.5)	240.0 (11.0)	299.0 (11.5)	159.3 (21.4)	185.7 (13.6)	152.0 (4.4)	163.3 (4.0)	254.3 (10.8)	191.0 (4.0)	149.3 (4.5)	137.0 (7.0)
5	334.3 (35.7)	171.7 (29.3)	269.3 (18.5)	261.7 (0.6)	209.7 (41.5)	174.0 (21.8)	107.0 (16.7)	243.3 (39.6)	333.7 (29.5)	339.7 (37.2)	229.0 (35.7)	192.3 (7.1)
6	176.3 (4.6)	177.7 (8.3)	287.0 (5.0)	266.0 (3.0)	204.3 (6.8)	110.3 (12.6)	136.7 (2.5)	182.7 (41.0)	248.7 (8.4)	229.0 (18.1)	161.7 (31.8)	161.3 (11.7)
7	173.3 (4.0)	140.0 (11.4)	183.3 (5.7)	206.0 (8.2)	148.0 (9.2)	259.3 (24.8)	292.7 (43.6)	203.3 (27.3)	143.0 (8.2)	135.0 (9.2)	96.7 (12.6)	125.3 (4.0)
8[a]	167.0 (1.7)	84.7 (22.0)	227.7 (15.0)	216.3 (13.1)	265.3 (40.5)	185.3 (15.9)	231.7 (3.1)	284.0 (54.8)	258.3 (2.9)	301.7 (15.8)	141.7 (10.8)	178.3 (11.6)
9[a]	178.5 (2.1)[b]	191.0 (13.9)[b]	306.5 (12.0)[b]	311.0 (39.6)[b]	166.0 (31.1)[b]	128.5 (4.9)[b]	119.5 (2.1)	168.5 (12.0)	326.5 (7.8)	322.5 (6.4)	190.5 (16.3)	183.3 (7.1)
mean (sd)	197.3 (54.5)	145.7 (32.2)	228.3 (51.1)	230.1 (56.1)	189.9 (56.1)	171.0 (60.8)	170.3 (79.0)	200.8 (46.7)	228.1 (74.7)	228.1 (76.1)	152.0 (39.9)	168.5 (30.4)

[a] pressure recordings not obtained from both feet simultaneously.
[b] average of 2 complete steps in the pressure recordings, nb. all other values are averages of 3 complete steps.

Table 5.11. Average (± 1sd) maximum peak pressures (kPa) beneath the feet of the patients during shod walking (sites of ulceration are underlined).

Patient	Left foot						Right foot					
	heel	mth4	mth3	mth2	mth1	hallux	hallux	mth1	mth2	mth3	mth4	heel
1	276.0 (4.0)	198.3 (41.8)	240.3 (32.3)	251.0 (42.8)	331.3 (68.7)	241.7 (4.9)	142.0 (24.2)	231.7 (36.7)	175.3 (15.2)	129.0 (7.8)	87.5 (34.9)	190.3 (6.8)
2	129.0 (28.3)[b]	208.0 (31.1)[b]	248.5 (41.7)[b]	259.5 (46.0)[b]	129.2 (45.0)[b]	123.5 (53.1)[b]	260.3 (62.3)	132.4 (42.5)	244.7 (64.9)	275.0 (46.2)	276.3 (40.6)	143.0 (38.2)
3[c]	166.7 (12.5)	148.3 (21.5)	220.3 (59.7)	199.3 (66.5)	262.7 (37.4)	139.3 (4.7)	-	-	-	-	-	-
4	180.7 (15.5)	156.0 (18.5)	252.3 (3.1)	363.0 (11.5)	228.3 (17.2)	173.3 (8.3)	92.0 (8.8)	150.3 (27.0)	231.7 (7.0)	177.3 (8.0)	117.3 (13.1)	179.3 (12.1)
5	293.7 (20.4)	116.1 (45.7)	164.3 (41.5)	152.7 (26.2)	319.3 (45.3)	145.0 (32.0)	36.9 (10.0)	261.7 (61.2)	213.7 (45.0)	173.7 (48.0)	115.4 (34.7)	217.7 (22.5)
6[d]	-	-	-	-	-	-	160.0 (46.7)	396.0 (236.2)	151.5 (40.3)	282.5 (62.9)	261.0 (83.4)	201.5 (20.5)
7[a]	245.0 (24.0)[b]	288.0 (0.0)[b]	231.5 (2.1)[b]	364.0 (17.0)[b]	269.5 (64.3)[b]	82.2 (15.3)[b]	206.0 (66.2)	380.0 (62.2)	504.3 (19.0)	376.3 (16.9)	362.3 (126.4)	204.3 (14.2)
mean (sd)[35]	224.9 (68.7)	193.3 (64.3)	227.4 (36.2)	278.0 (88.6)	255.5 (81.7)	153.1 (59.6)	147.4 (88.7)	231.2 (99.2)	273.9 (131.4)	226.3 (99.4)	191.8 (120.9)	186.9 (28.5)

[a] pressure recordings not obtained from both feet simultaneously.
[b] average of 2 complete steps in the pressure recordings, nb. all other values are averages of 3 complete steps
[c] pressure recordings were not obtained from beneath the right foot.
[d] pressure data not included (creasing of the F-Scan sensor inside the shoe made the pressure data unreliable).

Table 5.12. Correlations between local maximum peak pressures and age, and local maximum peak pressures and body mass, in the subject group.

Site of peak pressure	Age	Body mass
hallux	r = -0.204 (p=NS)	r = 0.114 (p=NS)
mth1	r = 0.753 (0.01<p<0.02)	r = 0.116 (p=NS)
mth2	r = 0.452 (p=NS)	r = 0.546 (p=NS)
mth3	r = 0.385 (p=NS)	r = 0.521 (p=NS)
mth4	r = 0.017 (p=NS)	r = 0.542 (p=NS)
heel	r = 0.131 (p=NS)	r = 0.282 (p=NS)

NS = not statistically significant at the 5% level, ie. p>0.05.

Table 5.13. Correlations between local maximum peak pressures and age, and local maximum peak pressures and body mass, in the patient group.

Site of peak pressure	Age	Body mass
hallux	r = -0.314 (p=NS)	r = 0.916 (0.02<p<0.05)
mth1	r = -0.222 (p=NS)	r = -0.206 (p=NS)
mth2	r = -0.157 (p=NS)	r = -0.033 (p=NS)
mth3	r = 0.111 (p=NS)	r = 0.311 (p=NS)
mth4	r = 0.152 (p=NS)	r = 0.432 (p=NS)
heel	r = -0.155 (p=NS)	r = -0.359 (p=NS)

NS = not statistically significant at the 5% level, ie. p>0.05.

For each complete gait cycle in the subject and patient pressure records, the timings of onset, maximum peak and cessation of pressure at local plantar sites were calculated as a percentage of the stance phase (Appendix E). Average timings were calculated by combining data from corresponding right and left foot sites. These values are presented in tables 5.14 and 5.15 for the subjects and patients, respectively; the calculated group means are presented pictorially in figure 5.4. For both groups the onset of local plantar pressures occurred in the same order: heel-mth4-mth3-mth2-mth1-hallux (figure 5.4). Local maximum peak pressures and the cessation of local pressures also occurred in this order in the subject group. In the patient group, local maximum peak pressures occurred in the order: heel-mth1-mth4-mth3-mth2-hallux; while the cessation of local pressures occurred in the order:heel-mth4-mth1-mth2-

mth3-hallux. There were no significant differences in the timings of onset, maximum peak and cessation of pressure at corresponding plantar sites beneath the feet of the subjects and patients (based on the data in tables 5.14 and 5.15[35]), with one exception: the onset of pressure beneath the hallux was much earlier into the stance phase in the subject group (p=0.0014). Consequently, there were no significant differences between the two groups in the duration of weight-bearing at the local plantar sites with the exception of the hallux. The subjects were weight-bearing longer on the hallux during walking (p=0.0015, unpaired t-test[35]).

Table 5.14. Average (± 1sd) timings of onset, maximum peak and cessation of pressure at local sites beneath the feet of the subjects during shod walking (expressed as a percentage of the stance phase).

Subject	heel			mth4			mth3			mth2			mth1			hallux		
	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation
1	0 (0)	19.7 (1.8)	52.0 (9.4)	6.7 (1.6)	62.8 (7.4)	95.1 (3.7)	11.3 (1.6)	76.3 (4.2)	97.3 (3.0)	12.1 (1.7)	77.8 (3.9)	98.3 (2.2)	11.3 (1.1)	74.4 (1.4)	99.5 (0.7)	33.5 (11.1)	76.1 (3.6)	97.1 (1.3)
2	0 (0)	22.3 (6.5)	60.8 (5.9)	18.9 (3.9)	71.7 (1.7)	89.6 (1.4)	20.6 (5.6)	75.1 (3.1)	90.4 (2.0)	25.5 (6.1)	75.9 (2.7)	90.1 (2.6)	48.5 (11.5)	81.6 (2.3)	97.2 (1.1)	13.5 (1.6)	88.4 (1.4)	100 (0)
3	0 (0)	21.3 (2.9)	46.9 (1.6)	5.3 (4.1)	71.3 (2.3)	99.4 (0.9)	7.5 (2.9)	74.3 (2.5)	99.4 (0.9)	14.8 (0.8)	77.5 (1.0)	99.4 (0.9)	18.2 (0.8)	77.5 (1.0)	98.8 (1.2)	6.0 (5.4)	85.2 (2.5)	100 (0)
4	0.6 (0.8)	21.0 (4.1)	60.8 (9.5)	19.0 (2.0)	75.3 (2.3)	97.2 (1.0)	20.2 (2.7)	77.8 (2.3)	97.8 (0.8)	22.2 (2.4)	80.6 (0.8)	96.9 (1.8)	31.7 (3.8)	80.1 (1.5)	96.3 (1.6)	18.5 (3.1)	87.7 (1.1)	99.4 (0.8)
5	0.3 (0.6)	20.3 (3.6)	88.4 (4.0)	9.2 (5.8)	73.0 (1.5)	99.8 (0.6)	13.4 (2.4)	76.2 (1.5)	99.5 (1.2)	14.6 (3.7)	76.9 (1.6)	100 (0)	22.5 (10.8)	79.2 (2.7)	99.5 (1.2)	28.8 (19.4)	83.9 (2.8)	99.5 (0.8)
6	0.6 (0.8)	9.1 (0.6)	55.6 (5.2)	16.9 (5.5)	81.3 (3.2)	98.8 (1.3)	18.4 (3.6)	82.8 (1.6)	99.4 (0.8)	18.1 (3.0)	81.0 (1.8)	99.7 (0.7)	15.0 (2.2)	77.8 (0.6)	100 (0)	57.8 (3.3)	87.8 (4.0)	99.4 (0.8)
7	0 (0)	15.4 (3.2)	68.2 (15.3)	13.5 (3.4)	73.7 (3.0)	95.6 (1.2)	13.5 (3.4)	76.0 (4.5)	96.6 (0.7)	14.3 (3.6)	78.6 (2.3)	97.6 (1.3)	18.1 (12.9)	82.8 (1.9)	98.7 (1.1)	28.7 (11.1)	85.7 (2.1)	99.7 (0.7)
8	0 (0)	19.3 (4.2)	65.0 (4.6)	11.9 (2.4)	72.8 (1.2)	98.6 (1.9)	10.5 (2.5)	77.7 (0.7)	99.7 (0.6)	13.6 (2.6)	79.9 (1.7)	99.4 (0.8)	17.1 (2.6)	79.5 (2.1)	99.7 (0.6)	12.4 (3.6)	86.9 (2.6)	100 (0)
9	0 (0)	23.4 (2.4)	82.1 (8.5)	9.9 (3.7)	76.8 (1.3)	98.4 (1.3)	9.3 (1.6)	80.7 (1.3)	100 (0)	10.0 (1.1)	82.0 (1.7)	100 (0)	12.8 (1.3)	80.0 (1.9)	99.7 (0.7)	33.6 (20.7)	88.4 (2.1)	98.1 (1.4)
mean (sd)	0.2 (0.3)	19.1 (4.4)	64.4 (13.5)	12.4 (5.1)	73.2 (5.0)	96.9 (3.2)	13.9 (4.8)	77.4 (2.7)	97.8 (3.0)	16.1 (5.0)	78.9 (2.1)	97.9 (3.1)	21.8 (11.7)	79.2 (2.5)	98.8 (1.3)	25.9 (15.6)	85.6 (3.9)	99.2 (1.0)

Table 5.15. Average (± 1sd) timings of onset, maximum peak and cessation of pressure at local sites beneath the feet of the patients during shod walking (expressed as a percentage of the stance phase).

Patient	heel			mth4			mth3			mth2			mth1			hallux		
	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation
1	0.6 (0.8)	25.2 (5.9)	71.8 (4.9)	26.5 (12.0)	77.1 (1.5)	89.2 (5.0)	30.8 (15.5)	78.9 (3.9)	93.6 (3.9)	32.9 (18.4)	80.7 (2.7)	95.9 (1.9)	35.5 (18.7)	81.5 (2.0)	99.1 (0.8)	57.5 (2.4)	90.6 (1.4)	99.1 (0.8)
2	0 (0)	16.5 (8.5)	57.1 (12.9)	11.9 (3.8)	72.7 (6.2)	91.4 (1.9)	13.7 (2.6)	75.9 (3.6)	94.9 (3.4)	29.0 (17.6)	81.1 (2.7)	94.6 (3.6)	34.2 (17.4)	65.4 (18.7)	90.1 (5.2)	52.3 (15.4)	90.3 (1.7)	99.1 (1.0)
3[a]	0.5 (0.8)	23.9 (6.2)	66.0 (5.9)	14.1 (3.2)	72.5 (4.0)	92.0 (2.1)	12.7 (1.7)	75.7 (0.7)	95.7 (1.8)	14.6 (3.6)	78.1 (2.9)	96.7 (0.8)	26.6 (20.3)	83.7 (2.3)	98.6 (1.4)	22.7 (5.2)	90.3 (1.8)	100 (0)
4	0.5 (0.7)	20.7 (6.8)	66.7 (6.2)	7.4 (4.1)	71.5 (7.9)	96.6 (1.6)	8.9 (2.8)	75.1 (4.3)	99.8 (0.5)	11.5 (6.2)	73.6 (8.2)	98.7 (1.5)	14.8 (4.5)	73.6 (4.8)	97.7 (1.6)	51.8 (34.8)	87.7 (1.6)	95.1 (2.8)
5	0.3 (0.6)	21.7 (4.0)	62.8 (5.0)	9.9 (1.0)	76.0 (1.7)	96.0 (2.2)	6.4 (0.7)	79.5 (1.1)	100 (0)	7.4 (1.5)	79.3 (0.7)	99.8 (0.6)	11.7 (4.5)	76.4 (2.2)	98.5 (1.3)	47.6 (6.0)	85.3 (1.7)	99.0 (1.2)
6[b]	0	14.3	67.3	3.1	32.7	92.9	7.1	75.5	92.9	7.1	78.6	92.9	13.3	78.6	99.0	45.9	84.7	94.9
7	3.1 (3.4)	18.1 (3.9)	58.0 (8.9)	12.4 (6.6)	73.0 (4.2)	93.1 (4.1)	12.6 (3.9)	78.0 (1.6)	99.1 (1.9)	16.3 (4.5)	76.8 (6.0)	98.2 (1.2)	19.0 (4.5)	62.6 (14.2)	98.3 (2.2)	42.3 (15.2)	79.3 (12.0)	97.9 (1.8)
mean (sd)[3]	0.9 (1.3)	20.4 (3.4)	63.3 (6.1)	13.6 (7.5)	74.1 (2.4)	93.3 (3.1)	14.5 (9.6)	77.5 (1.9)	97.5 (3.0)	19.4 (11.1)	78.3 (3.1)	97.4 (2.1)	23.0 (11.1)	71.9 (7.8)	96.7 (3.8)	50.3 (5.7)	86.6 (4.6)	98.0 (1.7)

[a] pressure recordings were not obtained from beneath the right foot - data for left foot only.
[b] creasing of the F-Scan sensor inside the shoe made the data from the left foot unreliable; there was only one complete gait cycle in the pressure recordings made with the right foot.

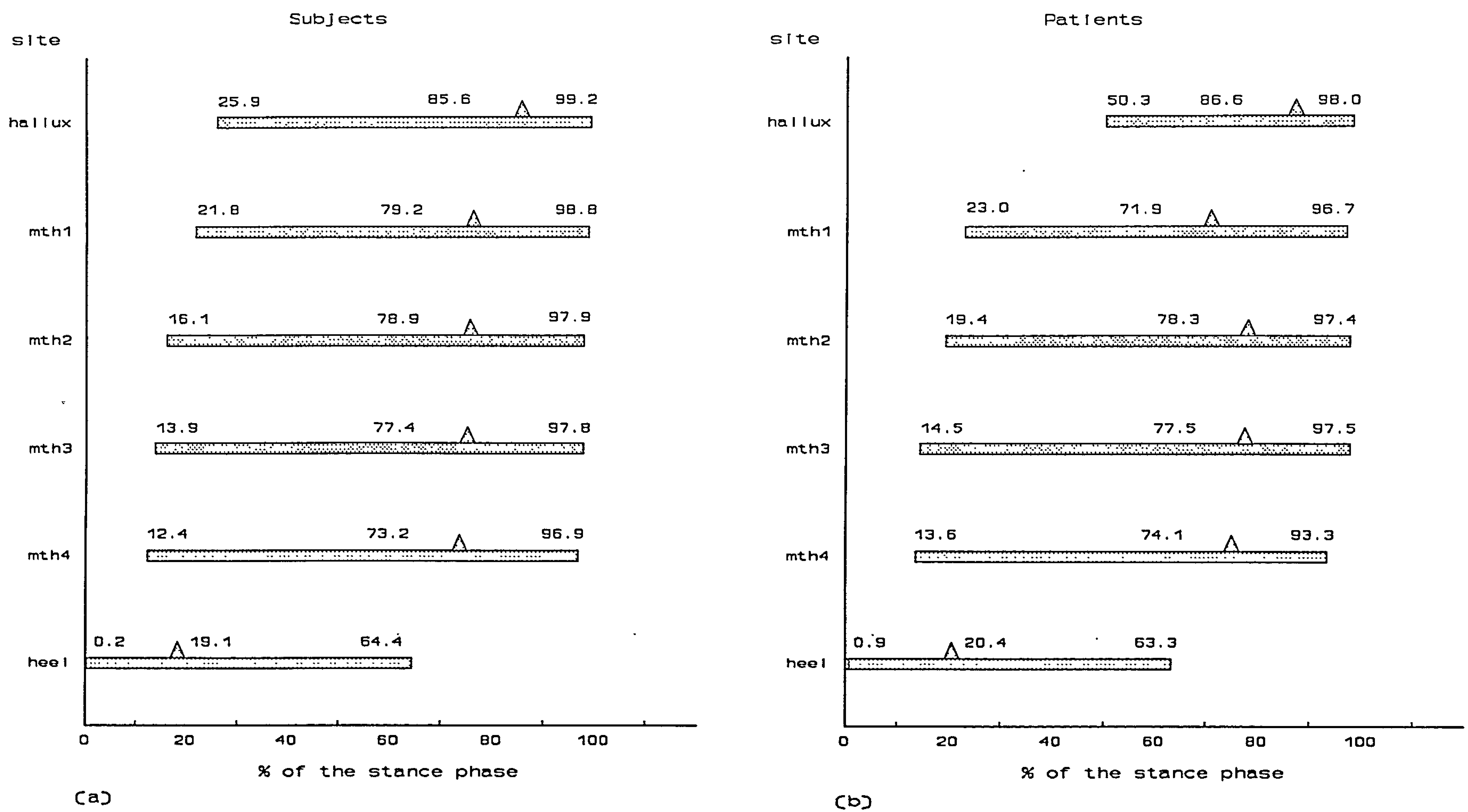

Figure 5.4. Average timings of the onset, peak (spikes) and cessation of pressure at local plantar sites expressed as a percentage of the stance phase.

5.4 SHEAR STRESS MEASUREMENT

Because of the number and complexity of the shear recordings which were made, it was decided before inspection to chose eight out of the sixteen records from each individual to minimise the analysis. These were the first recordings which were made with each inlay for each of the two conditions that were studied, ie. wearing and not wearing nylon hold-ups.

5.4.1 cadence

Footswitch data was not gathered from beneath both feet simultaneously while plantar shear stresses were being measured. As a result, it was not possible to determine the actual duration of a step for use in calculating the cadence. Instead, the duration of a step was estimated by dividing the duration of the first complete gait cycle in each of the shear records analysed (a total of eight for each individual) by a factor of two (tables 5.16 and 5.17). Unlike the analysis of the pressure recordings, an average cadence was not calculated for each shear record based on the duration of each complete step. In most cases, the duration of the first complete gait cycle was calculated by taking successive cues of heel-to-ground contact of the same foot from the footswitch data. In the few cases where the heel-switch was damaged during a test, the onset of longitudinal shear beneath the heel was used as the cue for *heel-to-ground contact* (see Appendices F and G for the subject and patient shear records, respectively). However, the shear stress data only provides an estimate of heel-to-ground contact. From an inspection of the shear records, it is apparent that the onset of longitudinal shear beneath the heel did not coincide with the closure of the heel-switch: there are cases where the onset of longitudinal shear occurred before the heel-switch closed and cases where the onset of longitudinal shear occurred after the heel-switch closed.

Subjects and patients walked at the same rate while in-shoe plantar shear stresses were being recorded ($p=0.16$, unpaired t-test[36], based on average data in tables 5.16 and 5.17). Even so, there was a greater variation amongst the patients compared to

[36] Data from patient number 3 was not included because shear stresses were not recorded from beneath both feet.

the subjects as indicated by the larger standard deviation of the group mean in table 5.17 compared with the group mean in table 5.16. Repeat shear tests, which were carried out on one subject and patient three months apart were analysed using the t-test to estimate differences between the two days. In the repeat shear tests, the subject walked at the same rate (p=0.053), whereas the patient walked slower (p=0.0081, paired t-tests). In a comparison between pressure and shear test data,the subjects walked at the same rate (p=0.54) as did the patients (p=0.90[35], paired t-tests). *Reproducibility* in the form of day-to-day variation, could not be determined because measurements were made on two days only.

Table 5.16. Average cadences during plantar shear stress measurements, for the subjects (steps/minute).

Subject	Cadence (left foot)				Cadence (right foot)				Average (sd)
	H-U 1-3-h	H-U 2-4-h	B 1-3-h	B 2-4-h	H-U 1-3-h	H-U 2-4-h	B 1-3-h	B 2-4-h	
1	107.6[a]	118.8[a]	115.9[a]	115.6[a]	117.6[a]	107.9[a]	119.8[a]	117.1[a]	115.0 (4.7)
2	107.6	107.0	107.4	105.9	110.9	109.9	108.2	112.3	108.7 (2.2)
3	111.1	105.4	106.2	107.4	108.2	108.1	109.3	105.7	107.7 (1.9)
4	97.2[a]	105.9[a]	102.6[a]	105.5[a]	108.3	101.8	109.4	95.5	103.3 (5.0)
5	105.4	107.4	107.1	108.2	109.9	107.1	111.8	104.8	107.7 (2.3)
6	114.1	113.0	115.1	113.2	113.9	114.3	112.5	117.6	114.2 (1.6)
7	114.4	115.9	118.8	118.8	115.6	117.1	118.2	118.6	117.2 (1.7)
8	112.7	119.2	111.1	119.2	108.6	111.3	113.4	109.1	113.1 (4.1)
9	101.9	94.1	103.4	103.7	102.6	100.1	101.4	98.2	100.7 (3.2)
mean (sd)									109.7 (5.6)
9-R	106.2	100.8	101.1	99.0	106.5	108.4	110.7	108.4	105.1 (4.3)

H-U 1-3-h - subject wearing nylon Hold-Ups, inlay instrumented with transducers beneath the 1st and 3rd metatarsal heads and heel.
H-U 2-4-h - subject wearing nylon Hold-Ups, inlay instrumented with transducers beneath the 2nd and 4th metatarsal heads and heel.
B 1-3-h - subject walking Barefoot in the shoes, inlay instrumented with transducers beneath the 1st and 3rd metatarsal heads and heel.
B 2-4-h - subject walking Barefoot in the shoes, inlay instrumented with transducers beneath the 2nd and 4th metatarsal heads and heel.
[a] cadences calculated based on the timing of onset of longitudinal shear beneath the heel.
R - results of the repeat shear test.

Table 5.17. Average cadences during plantar shear stress measurements, for the patients (steps/minute).

Patient	Cadence (left foot)				Cadence (right foot)				Average (sd)
	H-U 1-3-h	H-U 2-4-h	B 1-3-h	B 2-4-h	H-U 1-3-h	H-U 2-4-h	B 1-3-h	B 2-4-h	
1	107.4	113.9	104.6	99.6	97.4	103.7	97.0	107.0	103.8 (5.7)
2	100.0	95.4	109.4	114.3	91.8	99.0	101.9[b]	101.4	101.7 (7.2)
3[a]	111.1	115.9	119.4	111.6	-	-	-	-	114.5 (3.9)
4	92.9	78.5	73.8	86.8[b]	82.5	97.8	88.8	85.3	85.8 (7.7)
5	118.8	121.6	117.9	121.3	113.0	111.4	117.1	121.0	117.8 (3.8)
6	88.3	93.2	92.7	96.4	82.4	86.6	85.7	90.4[b]	89.5 (4.6)
7	101.1	105.7	103.4	106.5	98.6	105.3	100.3	107.4[b]	103.5 (3.2)
mean (sd)[c]									100.4 (11.5)
7-R	99.2	99.3	96.8	100.2	99.6	100.8	97.0	95.4	98.5 (1.9)

H-U 1-3-h - patient wearing nylon Hold-Ups, inlay instrumented with transducers beneath the 1st and 3rd metatarsal heads and heel.
H-U 2-4-h - patient wearing nylon Hold-Ups, inlay instrumented with transducers beneath the 2nd and 4th metatarsal heads and heel.
B 1-3-h - patient walking Barefoot in the shoes, inlay instrumented with transducers beneath the 1st and 3rd metatarsal heads and heel.
B 2-4-h - patient walking Barefoot in the shoes, inlay instrumented with transducers beneath the 2nd and 4th metatarsal heads and heel.
[a] shear stresses were not recorded beneath the right foot.
[b] cadences calculated based on the timing of onset of longitudinal shear beneath the heel."
[c] not including data from patient number 3, from whom shear stresses were only recorded beneath the right foot."
R - results of the repeat shear test."

5.4.2 shear stress records

The shear records for the subjects and patients are presented in Appendices F and G, respectively. Each Appendix is divided into two parts: Part I contains the shear stress data which was recorded while individuals were not wearing hosiery; while Part II contains the shear stress data which was recorded while individuals were wearing nylon hold-ups.

The shear records for an individual are presented on two pages, with one exception: longitudinal shear data is presented on the first of these pages and transverse shear data on the second. Because local plantar shear stresses were recorded beneath one foot of patient number 3, the longitudinal and transverse shear data for this patient appears on one page. Data from each foot appears on the corresponding side of each page, ie. right foot data on the right side.

Recall from Chapter 4, section 4.3.3, that two walks of three transducer variations were made for each foot under two conditions. Hence each record has ten separate traces. The top six traces are of either longitudinal or transverse shear stresses, which are plotted with respect to the direction of the shear force applied to the transducer; and the bottom four traces are of footswitch output. From top to bottom:

- traces 1 to 4 are of shear stresses generated beneath the first to fourth metatarsal heads, respectively - shear stresses were recorded simultaneously beneath the first and third metatarsal heads, and simultaneously beneath the second and fourth metatarsal heads;
- trace 5 is of the shear stress generated beneath the heel - recorded at the same time that shear stresses were recorded beneath the first and third metatarsal heads;
- trace 6 is of the shear stress generated beneath the heel - recorded at the same time that shear stresses were recorded beneath the second and fourth metatarsal heads;
- traces 7 and 8 are of heel- and toe-switch output, respectively - recorded at the same time that shear was recorded beneath the first and third metatarsal heads;

and, finally,

- traces 9 and 10 are of heel- and toe-switch output, respectively - recorded at the same time that shear was recorded from beneath the second and fourth metatarsal heads.

If the two walks to collect data from the medial four metatarsal heads were exactly repeatable then three pairs of traces would be identical in each shear record: traces 5 and 6, 7 and 9, and 8 and 10. Looking at traces 5 and 6 throughout the shear records, a remarkable level of consistency in walking is apparent.

Each trace in the shear records has been given a separation offset so that they are clearer to see. The vertical divisions on the graphs represents 50 kPa (for the shear stress data only). In a few records there is no heel- or toe-switch output, or else the output is erratic. In either case this is due to the footswitch becoming damaged during walking.

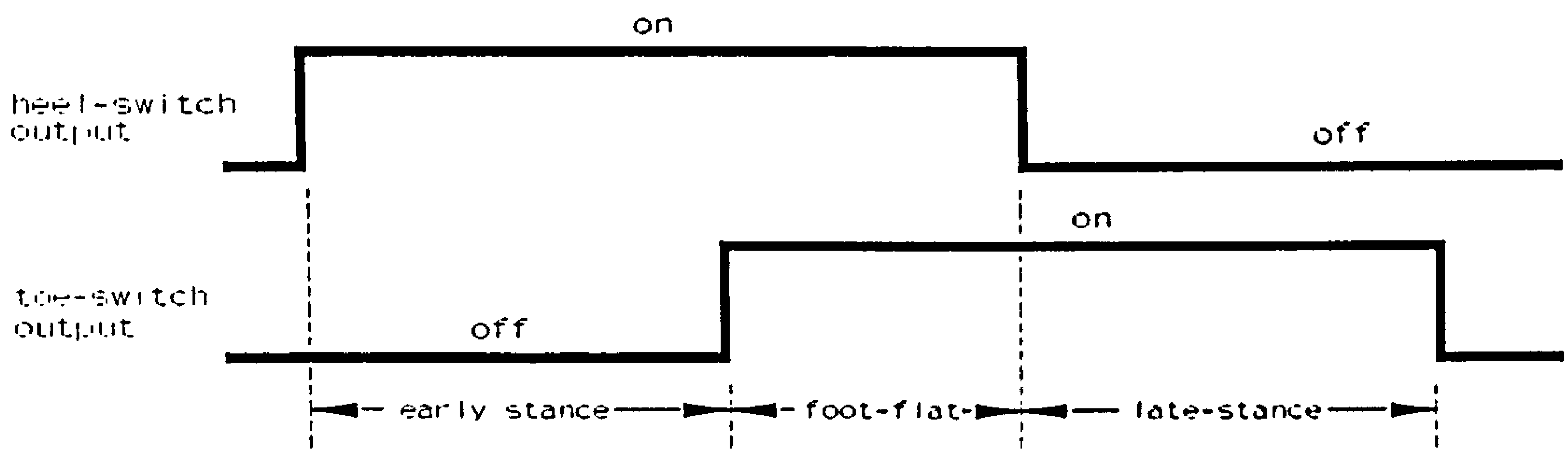

Figure 5.5. The three phases of gait that are defined with respect to the heel- and toe-switch outputs in the shear records.

Three phases of gait are defined with respect to the footswitch traces at the bottom of each record (figure 5.5). *Early stance* is defined as the period between the heel-switch going high (ie. at the instant when the heel-switch is closed) and the toe-switch going high. *Foot-flat* is defined as the period between the toe-switch going high and the heel-switch going low. Finally, *late stance* is defined as the period between the heel-switch going low and the toe-switch going low. The foot typically reaches a flat position as early as 7% into the gait cycle (**Mann, 1991**). Therefore it can be assumed from the footswitch output that the forepart of the shoe outsole made ground contact *before* the toe-switch was activated. In fact in some individuals there may have been a *significant* transfer of load to the forefoot before the toe-switch was activated.

5.4.3 maximum peak shear stresses

The magnitude of the *resultant maximum peak shear stress* measured beneath the medial four metatarsal heads and heel, for the first three complete gait cycles in the subject and patient shear records, are presented in Appendix H. Data is presented for the two conditions tested, ie. for individuals not wearing hosiery and wearing nylon hold-ups. The resultant maximum peak shear stress (*rmpss*) is calculated as a vector-addition of the longitudinal (*l*) and transverse (*t*) components of shear, ie:

$$rmpss=\sqrt{l^2+t^2} \tag{5.1}$$

For each individual, an average *rmpss* has been calculated for each site based on the data gathered from the first three complete gait cycles in the shear records. This data is presented in tables 5.18 to 5.21. In the subject group, the third metatarsal head was the site of highest *rmpss* beneath the forefoot followed by the first for both conditions tested (see tables 5.18 and 5.19). In the patient group, the first and second metatarsal heads were the sites of highest *rmpss* beneath the forefoot when not wearing hosiery (table 5.20); while the first metatarsal head was the site of highest *rmpss* followed by the second when wearing nylon hold-ups (table 5.21). It was noted that the sites of five out of nine healed plantar metatarsal ulcers were coincident with the sites of highest *rmpss* beneath the forefoot when not wearing hosiery (table 5.20); while the sites of six out of nine healed plantar metatarsal ulcers were coincident with the sites of highest *rmpss* beneath the forefoot when wearing hold-ups.

Table 5.18. Average (± 1sd) resultant maximum peak shear stresses (kPa) beneath the feet of the subjects during shod walking (not wearing hosiery).

Subject	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
1	12.3 (0.4)	20.0 (1.5)	31.3 (4.0)	60.1 (3.3)	34.5 (9.3)	51.1 (6.5)	26.3 (8.0)	15.7 (1.9)	43.1 (8.7)	37.2 (5.6)	40.1 (4.9)	26.8 (1.9)
2	34.1 (1.8)	33.4 (0.5)	37.0 (4.3)	31.2 (1.5)	32.4 (2.6)	56.5 (2.1)	70.8 (3.8)	22.9 (2.7)	30.6 (0.6)	36.3 (3.9)	29.3 (0.8)	26.9 (2.7)
3	12.7 (3.1)	13.2 (2.0)	34.0 (5.5)	27.7 (0.7)	47.2 (3.1)	42.6 (1.8)	9.4 (2.6)	8.5 (2.2)	37.1 (4.1)	28.2 (9.6)	16.1 (0.4)	22.0 (3.6)
4	41.5 (0.6)	46.0 (1.9)	51.9 (9.8)	107.0 (13.0)	20.8 (2.2)	51.0 (7.0)	41.9 (0.8)	32.4 (3.0)	114.6 (5.9)	112.1 (12.5)	51.2 (3.6)	48.2 (7.6)
5	30.0 (2.6)	19.9 (0.5)	44.2 (1.6)	51.7 (4.5)	27.2 (1.9)	14.6 (4.1)	23.6 (2.5)	10.8 (0.9)	108.2 (2.7)	69.8 (4.9)	42.9 (2.3)	41.7 (3.8)
6	118.3 (10.5)	97.4 (10.9)	24.3 (3.5)	37.7 (3.6)	19.3 (1.0)	38.5 (5.5)	32.8 (4.1)	14.2 (8.9)	58.5 (9.2)	44.2 (1.8)	43.6 (6.1)	81.2 (7.1)
7	45.7 (4.5)	47.3 (9.1)	36.5 (5.4)	45.7 (3.8)	27.0 (10.2)	43.9 (8.2)	51.4 (9.0)	61.2 (6.6)	104.3 (2.8)	132.7 (6.3)	88.6 (9.3)	78.0 (5.7)
8	45.0 (7.9)	85.5 (21.2)	25.3 (2.3)	61.9 (7.9)	30.1 (2.8)	50.0 (10.6)	28.7 (3.9)	49.6 (3.3)	86.7 (3.5)	78.4 (8.5)	78.9 (2.2)	67.3 (10.9)
9	52.8 (12.0)	39.1 (2.4)	59.7 (4.3)	30.6 (7.6)	115.6 (15.9)	69.5 (9.7)	29.1 (2.7)	63.4 (1.8)	195.7 (12.6)	100.5 (7.5)	45.4 (6.0)	40.1 (2.4)
mean (sd)	43.6 (31.4)	44.6 (29.2)	38.2 (11.8)	50.4 (24.8)	39.3 (29.8)	46.4 (14.9)	34.9 (17.8)	31.0 (21.8)	86.5 (52.1)	71.0 (37.6)	48.5 (22.6)	48.0 (22.5)
9-R	26.4 (4.7)	26.0 (1.9)	18.8 (1.7)	22.4 (4.6)	51.5 (4.0)	42.3 (4.2)	20.6 (3.5)	39.1 (3.1)	102.7 (6.9)	51.6 (7.1)	25.7 (1.8)	36.6 (4.5)

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
R - results of the repeat shear test.

Table 5.19. Average (± 1sd) resultant maximum peak shear stresses (kPa) beneath the feet of the subjects during shod walking (wearing nylon hold-ups).

Subject	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
1	34.9 (2.7)	22.1 (1.4)	13.1 (0.8)	24.1 (2.7)	15.7 (5.0)	33.7 (7.7)	10.5 (5.4)	11.4 (1.3)	35.3 (8.0)	44.2 (6.0)	22.2 (2.9)	23.1 (3.8)
2	30.9 (0.1)	33.7 (4.2)	32.7 (2.6)	32.2 (3.0)	46.9 (2.0)	48.1 (7.6)	37.7 (7.8)	13.9 (0.8)	36.6 (1.8)	40.1 (1.9)	23.9 (2.3)	22.4 (0.9)
3	18.6 (2.2)	19.9 (1.2)	15.5 (4.3)	20.0 (1.9)	18.5 (2.3)	12.4 (1.7)	18.9 (6.9)	15.6 (1.9)	24.8 (2.5)	37.9 (5.8)	9.1 (0.6)	14.1 (2.5)
4	33.8 (2.5)	43.9 (1.8)	33.0 (3.0)	102.2 (1.3)	28.9 (2.3)	75.2 (11.2)	50.1 (2.6)	33.0 (2.4)	100.9 (5.5)	72.7 (8.7)	53.9 (2.4)	49.7 (2.9)
5	32.6 (4.0)	29.5 (7.7)	28.1 (8.2)	62.3 (5.3)	32.9 (15.5)	11.1 (2.1)	29.6 (3.0)	14.1 (4.7)	91.2 (1.2)	65.6 (3.1)	37.6 (7.0)	34.3 (4.6)
6	79.8 (2.4)	58.9 (11.2)	21.2 (2.4)	45.3 (3.3)	28.3 (5.8)	43.6 (16.5)	39.1 (11.3)	11.0 (3.5)	47.6 (4.2)	31.4 (3.1)	73.4 (5.4)	67.4 (6.8)
7	49.9 (4.5)	42.4 (4.6)	26.9 (7.7)	44.2 (4.7)	29.9 (10.9)	104.3 (17.7)	92.3 (4.0)	31.1 (4.3)	60.7 (4.0)	79.6 (3.4)	115.1 (19.1)	90.2 (9.6)
8	52.6 (1.3)	76.0 (7.3)	20.9 (5.2)	66.9 (3.1)	30.7 (6.2)	65.2 (13.3)	23.7 (6.0)	52.7 (2.9)	111.6 (7.9)	57.4 (13.1)	97.9 (3.8)	85.1 (10.9)
9	43.2 (6.3)	44.8 (0.7)	41.1 (1.6)	38.6 (8.9)	81.2 (7.6)	23.5 (5.0)	44.5 (9.0)	32.9 (3.9)	124.9 (4.8)	79.5 (6.7)	29.5 (6.3)	39.3 (4.5)
mean (sd)	41.8 (17.6)	41.2 (17.9)	25.8 (9.0)	48.4 (25.5)	34.8 (19.5)	46.3 (30.9)	38.5 (23.8)	24.0 (14.3)	70.4 (37.3)	56.5 (18.7)	51.4 (36.7)	47.3 (27.9)
9-R	38.9 (0.6)	36.5 (1.9)	14.8 (1.3)	24.1 (5.3)	38.8 (5.3)	30.1 (3.4)	18.4 (3.9)	24.2 (4.9)	73.7 (6.1)	41.4 (4.3)	28.9 (3.5)	24.2 (3.9)

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
R - results of the repeat shear test.

Table 5.20. Average (± 1sd) resultant maximum peak shear stresses (kPa) beneath the feet of the patients during shod walking (not wearing hosiery - sites of previous ulceration beneath the forefoot are underlined).

Patient	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
1	84.4 (1.9)	61.0 (17.1)	53.1 (9.1)	64.9 (8.4)	54.1 (2.1)	63.5 (19.6)	121.8 (4.1)	98.9 (1.6)	52.5 (6.7)	46.6 (3.3)	52.4 (8.8)	70.5 (7.1)
2	9.0 (0.4)	12.7 (1.4)	39.2 (4.6)	46.2 (2.2)	52.1 (5.5)	15.8 (3.6)	50.8 (0.9)	72.5 (2.2)	62.1 (7.4)	45.1 (4.3)	41.4 (17.7)	45.9 (6.0)
3[a]	59.8 (4.3)	77.0 (6.6)	119.6 (19.7)	109.7 (7.7)	69.4 (26.2)	138.3 (31.4)	- -	- -	- -	- -	- -	- -
4	8.8 (1.5)	13.1 (4.0)	77.5 (6.5)	37.1 (2.3)	23.6 (3.6)	31.8 (14.6)	72.2 (24.9)	23.7 (6.1)	38.1 (12.6)	21.9 (1.1)	22.4 (0.7)	29.6 (2.9)
5	59.8 (4.9)	53.5 (3.3)	17.9 (5.7)	31.5 (2.5)	53.3 (1.3)	52.7 (3.5)	85.7 (7.5)	30.8 (7.8)	41.7 (1.7)	44.0 (1.6)	34.9 (4.0)	22.5 (2.7)
6	29.6 (3.2)	38.7 (10.6)	21.0 (2.9)	16.5 (2.7)	38.9 (6.2)	22.7 (4.2)	50.4 (13.0)	41.6 (3.9)	27.5 (2.5)	27.8 (2.9)	36.3 (10.4)	32.0 (2.6)
7	62.1 (13.5)	81.8 (9.9)	91.9 (24.8)	33.7 (1.4)	36.3 (2.9)	47.7 (7.3)	55.0 (4.8)	114.1 (3.6)	81.0 (8.8)	50.9 (4.0)	58.3 (15.2)	67.4 (8.2)
mean (sd)[b]	44.8 (29.2)	48.3 (28.1)	60.0 (38.0)	48.5 (30.8)	46.8 (15.0)	53.2 (41.2)	72.7 (27.8)	63.6 (37.5)	50.5 (19.1)	39.4 (11.7)	41.0 (12.9)	44.7 (20.3)
7-R	26.1 (5.3)	37.5 (6.4)	76.5 (3.0)	42.3 (1.3)	31.0 (2.0)	33.3 (8.8)	56.7 (8.2)	56.0 (13.0)	54.0 (12.5)	46.0 (4.8)	65.8 (7.2)	51.1 (4.3)

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
[a] shear stresses were not recorded beneath the right foot.
[b] data for the right foot of patient number 3 not included because plantar shear stresses were not recorded.
R - results of the repeat shear test.

Table 5.21. Average (± 1sd) resultant maximum peak shear stresses (kPa) beneath the feet of the patients during shod walking (wearing nylon hold-ups - sites of previous ulceration beneath the forefoot are underlined).

Patient	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
1	69.3 (0.3)	75.3 (1.1)	29.1 (3.0)	49.4 (10.0)	54.9 (15.9)	32.3 (3.2)	88.6 (3.8)	59.1 (8.9)	63.3 (2.2)	37.2 (14.7)	50.7 (5.0)	64.6 (7.2)
2	9.6 (1.1)	9.6 (3.9)	30.9 (4.1)	39.2 (6.3)	23.3 (2.3)	25.2 (2.3)	40.5 (12.3)	62.1 (4.1)	50.7 (13.6)	44.3 (2.9)	37.2 (5.0)	31.0 (7.9)
3[a]	46.7 (3.1)	53.9 (2.4)	89.7 (7.3)	102.0 (17.1)	110.5 (5.9)	135.9 (19.6)	- -	- -	- -	- -	- -	- -
4	19.7 (5.1)	8.7 (1.6)	80.7 (2.2)	38.2 (7.7)	15.8 (1.8)	21.3 (5.2)	58.3 (14.4)	35.9 (1.0)	45.6 (8.4)	33.6 (5.9)	22.6 (6.6)	34.5 (6.6)
5	64.6 (10.2)	53.9 (6.9)	14.8 (1.7)	35.6 (11.3)	68.6 (5.4)	33.8 (4.2)	45.3 (1.3)	22.4 (5.9)	43.4 (1.6)	40.1 (1.5)	34.4 (3.8)	18.2 (2.7)
6	34.5 (9.1)	33.7 (5.3)	17.5 (4.7)	46.3 (6.0)	39.6 (3.2)	36.1 (6.4)	53.1 (9.5)	44.7 (5.2)	29.1 (1.9)	36.4 (7.9)	31.5 (2.6)	31.2 (5.4)
7	60.1 (5.2)	83.3 (13.8)	62.2 (15.2)	41.3 (4.0)	36.0 (4.9)	64.6 (12.4)	59.9 (4.8)	121.4 (13.4)	51.3 (11.3)	42.0 (14.8)	48.3 (5.6)	54.4 (7.9)
mean (sd)[b]	43.5 (23.1)	45.5 (29.6)	46.4 (30.8)	50.3 (23.3)	49.8 (32.2)	49.9 (40.4)	57.6 (16.9)	57.6 (34.6)	47.2 (11.3)	38.9 (3.9)	37.5 (10.6)	39.0 (17.2)
7-R	35.3 (10.1)	43.0 (6.1)	61.4 (4.2)	59.8 (10.8)	60.8 (5.2)	47.9 (10.1)	51.9 (13.7)	63.3 (17.4)	74.6 (14.0)	39.4 (7.6)	38.3 (1.9)	54.4 (2.1)

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
[a] shear stresses were not recorded beneath the right foot.
[b] data for the right foot of patient number 3 not included because plantar shear stresses were not recorded.
R - results of the repeat shear test.

To check the consistency of the shear data, duplicate measurements of shear stress were made beneath the heel of each individual over the two walks routinely taken. For each group, for the two conditions tested, there was found to be no significant difference in the rmpss measured beneath the heel over the two walks (ie. $p>0.05$, paired t-tests).

For statistical analyses, the *rmpss* measured at corresponding sites beneath the left and right foot of each individual were averaged (data from tables 5.18 to 5.21). These averages were found to be similar between the two groups for each condition tested ($p>0.05$, unpaired t-tests); and between the two conditions tested for each group ($p>0.05$, paired t-tests) with one exception: the *rmpss* measured beneath the fourth metatarsal head in the subject group was greater when not wearing hosiery ($p=0.0062$).

The Pearson product moment (r) was calculated to correlate age and body mass with the local average *rmpss*. In the subject group, neither age or body mass was correlated; whereas in the patient group, there were several significant correlations. These are most conveniently listed:

- age vs rmpss measured beneath mth1, not wearing hosiery: r=-0.911, $0.0001<p<0.01$
- age vs rmpss measured beneath the heel, not wearing hosiery: r=-0.789, $0.02<p<0.05$
- age vs rmpss measured beneath the heel, wearing nylon hold-ups: r=-0.827, $0.02<p<0.05$
- body mass vs rmpss measured beneath mth2, not wearing hosiery: r=0.785, $0.02<p<0.05$
- body mass vs rmpss measured beneath mth3, not wearing hosiery: r=0.864, $0.01<p<0.02$
- body mass vs rmpss measured beneath mth3, wearing nylon hold-ups: r=0.780, $0.02<p<0.05$

Measurements of plantar shear were carried out on one subject and one patient three months apart to estimate differences in the magnitude of local *rmpss* over time. In a comparison of the data from each individual, seen in tables 5.18 to 5.21, obvious differences in average *rmpss* between the two tests are apparent. Although it is preferable to make measurements on several more occasions to assess reproducibility in a statistical analysis, it was not possible for the subject or patient to make multiple visits.

For each group, a comparison was made between the plantar pressure and shear stress data recorded beneath the medial four metatarsal heads and heel. This comparison revealed average maximum peak pressures were significantly higher than average resultant maximum peak shear stresses in each group, for each condition tested (ie. p<0.001, unpaired t-test).

5.5 BIOMECHANICAL EXAMINATION

5.5.1 repeatability

Intra-observer examinations were carried out on two subjects on four consecutive days (24 hours apart) to assess repeatability. The results are presented in tables 5.22 and 5.23 (for subjects number 3 and 9, respectively). The reference lines drawn on the skin of the foot and leg at the beginning of each examination were washed off by the subject before the next examination.

Inconsistencies were expected in the measurement of ranges of joint motion, particularly of the hindfoot, in line with other workers (**Elveru et al, 1988b**), but these were still higher than expected (see coefficients of variation and standard deviations for left and right foot calcaneal inversion in tables 5.22 and 5.23). These inconsistencies may be due to errors in locating and marking the reference lines on the skin of the foot and leg and/or in aligning the goniometer on these lines or may be due to variations in the levels of activity of the subjects prior to the examinations, which may have caused stretching of the muscles in the lower limb to a greater or lesser extent. To minimise the levels of activity prior to the examination a fixed policy was maintained throughout the four days of repeat examinations: the subjects arrived by the same mode of transport on each occasion and the examinations were conducted on arrival (subject number 9) or an hour after (subject number 3 reported late after arrival for the first appointment).

Intra-observer assessments of frontal plane deviations of the rearfoot and tibia were highly reliable in subject number 9: coefficients of variation indicate these measures were accurate to within ± 1.4% (see table 5.22). In subject number 3 this situation was altered. Repeatability for these same assessments was lower to begin with, being 75% for the assessment of tibial deviation in both legs and rearfoot deviation in the right foot. The reliability in assessing the ranges of motion at the ankle and subtalar joints was poor. In the worst case the measurement of ankle

Table 5.22. Intra-observer results of the repeat biomechanical examination carried out on subject number 3.

Joint motion/frontal plane deviation	Left foot test number 1	2	3	4	mean ± 1sd	coefficient of variation	Right foot test number 1	2	3	4	mean ± 1sd	coefficient of variation
ankle dorsiflexion	6°	11°	18°	11°	11.5° ± 4.9°	42.6%	13°	12°	9°	12°	11.5° ± 1.7°	14.8%
calcaneal inversion calcaneal eversion	28° 12°	30° 14°	30° 12°	26° 14°	28.5° ± 1.9° 13.0° ± 1.2°	6.7% 9.2%	20° 15°	37° 16°	18° 12°	20° 16°	23.8° ± 8.9° 14.8° ± 1.9°	37.4% 12.8%
forefoot varus forefoot valgus	 b	 b	 b	 b			 b	a a	 b	 b		
dorsiflexed 1st metatarsal plantarflexed 1st metatarsal	✓	a a	✓	✓			✓	✓	✓	✓		
dorsiflexed 5th metatarsal plantarflexed 5th metatarsal	✓	✓	✓	✓			 ✓	✓	✓	✓		
tibial varum tibial valgum	96°	94°	 87°	98°	- -	- -	96°	96°	 86°	92°	- -	- -
rearfoot varus rearfoot valgus	97°	98°	97°	97°	97.3° ± 0.5°	0.5%	a a	91°	93°	92°	- -	- -
angle of gait	4°	18°	5°	6°	8.3° ± 6.6°	79.5%						

[a] condition or frontal plane deviation was not present.
[b] total-type forefoot valgus.

Table 5.23. Intra-observer results of the repeat biomechanical examination carried out on subject number 9.

Joint motion/frontal plane deviation	Left foot test number 1	2	3	4	mean ± 1sd	coefficient of variation	Right foot test number 1	2	3	4	mean ± 1sd	coefficient of variation
ankle dorsiflexion	12^0	10^0	9^0	12^0	$10.8^0 \pm 1.5^0$	13.9%	10^0	10^0	9^0	7^0	$9.0^0 \pm 1.4^0$	15.6%
calcaneal inversion calcaneal eversion	20^0 9^0	25^0 9^0	21^0 8^0	20^0 13^0	$21.5^0 \pm 2.4^0$ $9.8^0 \pm 2.2^0$	11.2% 22.4%	12^0 7^0	22^0 8^0	24^0 7^0	19^0 9^0	$19.3^0 \pm 5.3^0$ $7.8^0 \pm 1.0^0$	27.5% 12.8%
forefoot varus forefoot valgus	[b]	[b]	[b]	[b]			[b]	[b]	[b]	[b]		
dorsiflexed 1st metatarsal plantarflexed 1st metatarsal	✓	✓	[a] [a]	[a] [a]			✓	✓	✓	✓		
dorsiflexed 5th metatarsal plantarflexed 5th metatarsal	 ✓	✓	✓	✓			✓	✓	[a] [a]	✓		
tibial varum tibial valgum	100^0	100^0	101^0	99^0	$100^0 \pm 0.8^0$	0.8%	94^0	94^0	95^0	94^0	$94.3^0 \pm 0.5^0$	0.5%
rearfoot varus rearfoot valgus	94^0	95^0	97^0	96^0	$95.5^0 \pm 1.3^0$	1.4%	92^0	92^0	92^0	91^0	$91.8^0 \pm 0.5^0$	0.5%
angle of gait	15^0	8^0	12^0	11^0	$11.5^0 \pm 2.9^0$	25.2%						

[a] condition or frontal plane deviation was not present.
[b] total-type forefoot valgus.

dorsiflexion was accurate to within ± 42.6% (subject number 3, left foot). The assessment of frontal plane deviations of the forefoot was highly repeatable in subject number 9 (100%); while in subject number 3, repeatability for this assessment was different for each foot: 100 % for the left foot and 75% for the right. Repeatability in the assessment of first and fifth metatarsal mobility was either 100% or 75% in subject number 3; similarly in subject number 9 with one exception: there was a 50% repeatability for the assessment of first metatarsal mobility in the left foot. The reliability in measuring the angle of gait was very poor. In the worst case this measurement was accurate to within ± 79.5% (subject number 3).

To summarise: during the biomechanical examination, angular ranges of motion at joints in the hindfoot, frontal plane deviations of the rearfoot and tibia, and the angle of gait were measured; and first and fifth metatarsal mobility was assessed. The few repeat examinations that were carried out did not provide consistent data so it is difficult to say outright which measures/assessments are the most repeatable - it is obviously necessary to study many more subjects in order to do so. Intuitively it is thought the assessments would be the most repeatable because they do not include the measurement of angles. However, having said this, even the assessments may be variable from one examination to the next depending on the amount of subject activity beforehand, which would affect the tension in the muscles of the foot and leg. Fortunately it is easy to control for this by instructing the subject not to perform strenuous exercises prior to arriving for the examination or by allowing a period of rest on arrival.

5.5.2 ranges of joint motion and frontal plane deviations of the foot and leg

Ranges of joint motion and frontal plane deviations in the feet and legs of the subjects and patients are presented in tables 5.24 and 5.25, respectively. The findings of the biomechanical examination are listed below:

- the range of ankle dorsiflexion was greater than or equal to 12^0 (the value considered sufficient for normal gait) in six out of the eighteen feet in the subject group (33%) and in two out of the twelve feet in the patient group

(17%)[37];

- the measured ranges of ankle dorsiflexion were similar between the groups (subject mean: $9.4^0 \pm 6.3^0$; patient mean: $8.4^0 \pm 2.2^0$, p=0.66, unpaired t-test, nb. all statistical tests were conducted on data averaged from both feet);
- in all subjects and patients the range of calcaneal inversion was lower than 36.8^0 (the value considered sufficient for normal gait);
- the measured ranges of calcaneal inversion were similar between the groups (subject mean: $22.7^0 \pm 7.7^0$; patient mean: $19.9^0 \pm 5.7^0$, p=0.43, unpaired t-test);
- in all subjects and patients the range of calcaneal eversion was lower than 20.7^0 (the value considered sufficient for normal gait);
- the measured ranges of calcaneal eversion were similar between the groups (subject mean: $10.9^0 \pm 2.6^0$; patient mean: $8.3^0 \pm 4.4^0$, p=0.23, unpaired t-test);
- the measured ranges of calcaneal inversion were more than double the ranges of calcaneal eversion in both groups as expected (subjects: $22.7^0 \pm 7.7^0$ and $10.9^0 \pm 2.6^0$, respectively, p=0.0019; patients: $19.9^0 \pm 5.7^0$ and $8.3^0 \pm 4.4^0$, respectively, p=0.0033), however, in both groups there were two individuals who were assessed to have an unexpectedly greater range of calcaneal eversion than calcaneal inversion in one or both feet;
- the measured ranges of ankle dorsiflexion, calcaneal inversion and calcaneal eversion were similar between the contralateral feet of the subjects (p=0.39, p=0.53 and p=0.89, respectively) and the patients (p=0.28, p=0.14 and p=0.58, respectively, paired t-tests).
- a forefoot valgus condition was observed in almost every individual (subjects: right foot 78%, left foot 89%; patients: both feet 83%);
- in both groups a dorsiflexed first metatarsal was more common than a plantarflexed first metatarsal in the right foot (subjects: right foot 89%, left foot 44%; patients: right foot 67%, left foot 33%);
- in both groups a dorsiflexed fifth metatarsal was more common than a plantarflexed fifth metatarsal in the left foot (subjects: right foot 33%, left foot 67%; patients: right foot 17%, left foot 50%);

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[37] The data from patient number 3 was not included in any of the analyses in this section because a biomechanical examination was not carried out on the right foot (see section 5.2).

Table 5.24. Ranges of joint motion and frontal plane deviations of the foot and leg in the subject group.

Joint motion/frontal plane deviation	Left foot Subject 1	2	3	4	5	6	7	8	9	Right foot Subject 1	2	3	4	5	6	7	8	9
ankle dorsiflexion	12^0	3^0	8^0	2^0	4^0	2^0	26^0	15^0	5^0	14^0	2^0	11^0	10^0	15^0	5^0	21^0	7^0	8^0
calcaneal inversion	31^0	30^0	21^0	34^0	36^0	11^0	18^0	27^0	6^0	26^0	28^0	26^0	19^0	18^0	6^0	24^0	31^0	17^0
calcaneal eversion	18^0	12^0	11^0	9^0	7^0	13^0	6^0	10^0	13^0	10^0	10^0	11^0	7^0	14^0	18^0	8^0	10^0	9^0
forefoot varus		a									a						a	
forefoot valgus	c	a	b	b	b	c	b	b	b	c	a	b	b	b	c	b	a	c
dorsiflexed 1st metatarsal					a	✓	✓	✓	✓	a	✓	✓	✓	✓	✓	✓	✓	✓
plantarflexed 1st metatarsal	✓	✓	✓	✓	a					a								
dorsiflexed 5th metatarsal	✓	a	✓	✓	✓	a	✓		✓			✓		✓	a	✓		
plantarflexed 5th metatarsal		a				a		✓		✓	✓		✓		a		✓	✓
tibial varum	92^0	102^0	92^0	94^0		a	92^0	96^0	104^0	99^0	98^0	a	102^0	a	a	92^0	97^0	
tibial valgum					88^0	a						a		a	a			80^0
rearfoot varus	92^0	a	a	91^0	a	96^0		92^0	96^0	a	a	98^0	94^0	a	96^0		92^0	
rearfoot valgus		a	a		a		87^0			a	a			a		87^0		80^0
angle of gait	6^0	60^0	21^0	22^0	16^0	18^0	13^0	22^0	12^0									

[a] condition or frontal plane deviation was not present.
[b] total-type forefoot valgus.
[c] plantarflexed first-metatarsal-type forefoot valgus.

Table 5.25. Ranges of joint motion and frontal plane deviations of the foot and leg in the patient group.

joint motion/frontal plane deviation	Left foot							Right foot						
	Patient							Patient						
	1	2	3	4	5	6	7	1	2	3[d]	4	5	6	7
sites of healed ulcers	mth1		mth1	mth1,5	mth2/3	mth1	mth1	mth1	mth2					mth2
ankle dorsiflexion	8^0	8^0	5^0	8^0	2^0	6^0	4^0	9^0	8^0	-	2^0	20^0	13^0	8^0
calcaneal inversion	21^0	7^0	15^0	7^0	19^0	18^0	18^0	31^0	21^0	-	18^0	29^0	20^0	30^0
calcaneal eversion	4^0	8^0	5^0	11^0	9^0	8^0	14^0	3^0	2^0	-	3^0	19^0	5^0	13^0
forefoot varus						✓			✓	-				
forefoot valgus	c	b	c	c	b		b	c		-	c	b	c	b
dorsiflexed 1st metatarsal					✓	✓		✓		-		✓	✓	✓
plantarflexed 1st metatarsal	✓	✓	✓	✓			✓		✓	-	✓			
dorsiflexed 5th metatarsal	✓	✓	✓		a		✓	✓		-				a
plantarflexed 5th metatarsal				✓	a	✓			✓	-	✓	✓	✓	a
tibial varum	92^0	100^0	120^0			96^0	91^0	94^0	94^0	-			102^0	a
tibial valgum				86^0	80^0					-	88^0	84^0		a
rearfoot varus	98^0		114^0		92^0		95^0	92^0	96^0	-	95^0	93^0	96^0	
rearfoot valgus		88^0		86^0		86^0				-				89^0
angle of gait	37^0	19^0	73^0	28^0	25^0	30^0	14^0							

[a] condition or frontal plane deviation was not present.
[b] total-type forefoot valgus.
[c] plantarflexed first-metatarsal-type forefoot valgus.
[d] a biomechanical examination was not carried out on the right foot of this patient.

- in both groups a tibial varum condition was more common than a neutral tibial position or a tibial valgum (subjects: right foot 56%, left foot 78%; patients: right foot 50%, left foot 67%); and a rearfoot varus condition was more common than a neutral calcaneal position or rearfoot valgus (subjects: right foot 44%, left foot 56%; patients: right foot 83%, left foot 50%);

and

- the angle of gait was similar between the groups (subject mean: $21.1^0 \pm 15.5^0$; patient mean: $35.3^0 \pm 19.4^0$, p=0.17, unpaired t-test).

Joint mobility was noticeably reduced in the hindfoot of the patients: a larger proportion of patients had below normal ranges of motion at the ankle and subtalar joints. In the forefoot, the frequency with which a valgus condition was seen in both groups, especially amongst the subjects, suggests this condition may be the norm in the general population. The assessment of first and fifth metatarsal mobility arouses some suspicion because of the frequency of a dorsiflexed first metatarsal in the right foot of both groups and a dorsiflexed fifth metatarsal in the left foot of both groups. It is possible that the handedness of the examiner was influential in these assessments. The examiner in this study was right-handed and used the left hand to dorsiflex and plantarflex these metatarsals. It is possible that compensation for the weakness of the left hand resulted in the application of excessive force.

It is particularly interesting to note in the patient group, that five out of the seven healed plantar ulcers beneath the first or fifth metatarsal heads were coincident with the respective metatarsal being plantarflexed; and five out of the six beneath the first metatarsal head were coincident with a forefoot valgus condition.

CHAPTER 6
DISCUSSION:
PLANTAR PRESSURE MEASUREMENT

6.1 INTRODUCTION

The transfer of bodyweight through the weight-bearing leg to the ground during walking generates a time-varying pattern of plantar pressure. In the present study, these patterns were measured at the foot/shoe interface of two groups: asymptomatic subjects wearing standard extra-depth orthopaedic shoes and patients with diabetes wearing their own bespoke shoes.

The pressure patterns recorded were primarily analysed to locate sites of peak pressure beneath the metatarsal heads and heel so that discrete shear transducers could be located in regions of maximum load associated with these anatomical sites (**Lord et al, 1992**). Other analyses were performed on the pressure data in an investigation of shod foot function. This Chapter discusses the design and technical aspects of this study; compares the subject data presented in Chapter 5 with the data from other barefoot and in-shoe plantar pressure studies; and compares the subject data to the patient data.

6.2 APPRAISAL OF THE CLINICAL TRIAL

Plantar pressures are generated during the stance phase by the bony anatomy of the foot acting through the plantar soft tissues. Therefore on an individual basis the thickness of the plantar soft tissue may be important in determining the time-varying patterns and magnitudes of plantar pressure that are recorded. However, apart from individual factors, the patterns and magnitudes of plantar pressure recorded may be affected by *study-design* related factors. Hence as a basis to a discussion of the plantar pressure data recorded, consideration will first be given to the different characteristics of the two groups studied; the limitations of the methods and measuring equipment employed; and the nature of the quantity being measured, which can be highly variable if not measured under the imposition of standardised conditions.

6.2.1 study groups and methods

So as to avoid delaying the completion of the clinical trial, the two groups studied were not matched in numbers or for age, body mass and sex. With regard to the methods employed: the shoes worn by the individuals of both groups were matched, although the patients did not wear their normal cushioning or moulded shoe inserts; and a controlled walking speed was not imposed during measurements.

A strict selection criteria was imposed in order to recruit suitable patients to investigate the role of plantar pressure and shear stress in causing ulceration of the insensitive diabetic foot. As a result, considerable difficulty was encountered in patient recruitment and only a small number were eventually studied. A history of plantar ulceration was the main criteria for patient selection. The main criteria for subject selection was foot size: subjects were required to fit into the particular sizes of extra-depth stock orthopaedic shoes that were used throughout.

The patients recruited to the trial were older and heavier than the subjects. Several studies have found age (**Hughes et al, 1990**) and body mass (**Soames, 1985; Cavanagh et al, 1991a; Snow et al, 1992; van der Zande et al, 1992**) to be poorly correlated with the magnitudes of local maximum peak pressures beneath the feet of healthy subjects. However, degeneration of the soft tissue beneath the metatarsal heads, similar to the heel pad with ageing (**Jorgensen and Bojsen-Moller, 1989**), may result in an increase in local maximum peak pressure beneath the metatarsal heads. High resolution ultrasonography is perhaps the simplest method of determining soft tissue thickness, but the necessary equipment for such an investigation was not available for use during this study.

There was a predominance of men in the two groups studied. This was purely due to chance in the patient group where consecutive patients were recruited who fulfilled the selection criteria. In the subject group, the large sizes of extra-depth stock orthopaedic shoes that were available predisposed towards male subjects. Both **Soames**

(1985) and **Holmes et al (1991)** found no significant difference in the local maximum peak pressures beneath the feet of men and women indicating no gender differences. Because of this and the fact that the groups in the present study were small with very few women in each, the data from both sexes in each group was combined for analysis.

Subjects wore a similar type of shoe to the patients so that there could be a more valid comparison of plantar pressures between the two groups. Shoe inserts were also standardised by replacing the patients' normal cushioning or moulded shoe insert with a flat inlay. Patients tend to ulcerate while wearing fashion shoes with a flat insole and are typically supplied with bespoke shoes and cushioning or moulded inserts at King's College Hospital only *after* they have first ulcerated. The present study was designed to measure local plantar stresses prior to the provision of special shoewear, which is intended to reduce local peak pressures and peak shear stresses thought to be risk factors in causing ulceration.

Many workers have favoured a controlled walking speed during pressure measurements (for example **Holmes et al, 1991; Zhu et al, 1991; Snow et al, 1992; van der Zande et al, 1992**), but this was not desirable here. A controlled walking speed is known to result in step-to-step variation in the timing of the stance phase (**Hughes et al, 1991b**), which may be accentuated over a short distance: the length of the trailing cable connecting the pressure sensor to the PC (9.5 m) restricted individuals to a short walk - a maximum of 19 m. As a result, individuals were allowed to walk at their own speed during measurements. The influence of walking speed on the magnitudes of peak pressure beneath the foot was studied by **Clarke** (cited in **Hennig and Rosenbaum, 1991**), who found an increase in walking speed from 1.33 ms^{-2} to 1.79 ms^{-2} resulted in an average increase of only 7.2% in the magnitudes of local peak pressures beneath all regions of the foot. The speed of walking was not measured during this study as there was no accurate and repeatable method available for doing this. However, there was no significant difference in cadence between the groups, which asserts that individuals of both groups walked at the same rate during pressure measurements.

6.2.2 equipment

The pressure sensor used in the present study is based on force-sensitive resistive technology (FSR). FSR technology represents a significant advance in plantar pressure measurement, which allows a thin, high resolution sensor to be fabricated for placement inside the shoe. However, when using these insole sensors it is important that they are 'worn-in' and equilibrated to the temperature inside the shoe before calibration in order to optimise the reliability of the measurements obtained.

The F-Scan[8] sensors used in the present study are not pre-conditioned by the manufacturer. A new sensor is highly sensitive, but over the first few gait cycles this gradually decreases and levels off (manufacturers literature). This is thought to be due to microscopic surface irregularities of the conductive ink tracks, but as yet untested: extensive trials of the time-varying properties of the sensors are being conducted by other members of this Department, but the results are not yet available. In the present study, an individual completed at least the twelve gait cycles recommended by the manufacturer, with the sensor in the shoe, before calibration and recordings were made. Although this wearing-in procedure conditions the parts of the sensor that will record plantar pressure, it may do so in an uneven manner. Hence the recommended wearing-in procedure may result in some parts of the sensor being more sensitive to pressure than others.

During the period of pre-conditioning, the sensor was also equilibrated to the temperature inside the shoe. In another study that was conducted using an earlier prototype version of the F-Scan system, the sensor was more sensitive to temperature. However, a 3-minute period of inactivity with the sensor inside the shoe was found to be sufficient time under the environmental conditions to achieve temperature equilibration **(Lord and Hosein, 1994).**

The F-Scan sensors were statically calibrated against the bodyweight of the individual being tested, in accordance with the method recommended by the manufacturer. However, when using the sensors to make dynamic measurements it is important to know their dynamic behaviour and to calibrate them under dynamic conditions, similar to those that will be encountered during measurements. Unfortunately, the necessary equipment for such dynamic calibration was not available for use during this study.

The manufacturer's recommended method of sensor calibration produces a unique conversion constant that assumes all loadcells are of equal sensitivity. However, personal communication with the manufacturer has revealed cell-to-cell variation may be as high as 50%, although this is somewhat reduced by a smoothing algorithm invoked by an *averaging* function in the software. Furthermore, the manufacturer states the errors that are introduced into measurements by their recommended method of calibration will only be a small percentage of the large pressures being recorded.

The pressure threshold of the sensor was investigated and found to be between 20 kPa and 30 kPa. It is thought unlikely that this threshold level compromises the sensor to measure pressures accurately over 30 kPa, but the start and finish of the locus of the centre of pressure may not be accurately determined, nor the onset and cessation of local plantar pressures accurately timed. It should also be noted that the errors introduced in determining the locus of the CoP and into the measurement of contact times, as a result of the threshold level of the sensor, will be compounded by a slow data sampling rate. During the present study this was compensated for by using a data sampling rate of 100 Hz throughout.

In tests with an earlier version of the F-Scan system it was noticed that the foot stuck to the sensor and dragged it forward when the shoe was put on. This resulted in creasing of the sensor with a subsequent loss of data. In order to avoid this problem in the present study, individuals wore nylon hold-ups. Even so, there was one case where a sensor creased inside the shoe of a patient due to a particularly shallow angle of heel-to-ground contact. This creasing made any analysis of the magnitudes of local peak pressure and the locus of the CoP unreliable, although it was possible to analyse the pressure record for *sites* of local peak pressure beneath the forefoot.

6.3 COMPARISON OF THE NORMATIVE IN-SHOE PLANTAR PRESSURE DATA GATHERED DURING THE CLINICAL TRIAL AND DATA FROM BAREFOOT PLANTAR PRESSURE STUDIES

The normative in-shoe plantar pressure data that was gathered in the present study is compared below against the findings of workers who measured barefoot plantar pressures. The basis for this comparison was to determine whether alterations in foot function as a result of wearing lace-up shoes could be determined from the pressure data that was gathered during the clinical trial.

6.3.1 locus of the centre of pressure

Seen against an outline of the foot, the locus of the centre of pressure (CoP) in-shoe that was observed in the subject group deviated from the typical locus seen in barefoot walking in three important respects (see table 5.8). Firstly, the locus of the CoP in-shoe originated beneath the centre of the heel as opposed to the postero-lateral aspect; secondly, it progressed forward beneath the midfoot just lateral to the midline as opposed to beneath the lateral border; and thirdly, it travelled beneath the centre of the forefoot as opposed to traversing from the lateral to medial side. Similarities between barefoot and shod walking were noted with the locus of the CoP in-shoe terminating beneath the lateral aspect of the hallux.

The locus of the CoP in-shoe was expected to originate beneath the postero-lateral aspect of the heel, similar to barefoot walking. As in barefoot walking, the shod foot typically approaches the ground in an inverted position at the end of the swing phase and as a result the postero-lateral border of the heel is usually the first part of the shoe to make contact with the ground at the beginning of the stance phase. The pressure threshold of the sensor used in the present study may have significantly influenced the location of the origin of the CoP in-shoe beneath the centre of the heel. In most cases, the onset of pressure was beneath the centre of the heel (as indicated by the origin of the locus), implying the heel of the shoe came down vertically onto the ground at heel-contact. From observing individuals during pressure measurements it was obvious that this was not the case. It is probable that the postero-lateral aspect of the heel *was* the first part of the shoe to make ground contact, but pressure generated beneath the postero-lateral aspect of the heel pad of the foot was below the 20 kPa threshold of the insole sensor to be registered. It appears that compression of the heel pad redistributed pressure to below the threshold of the F-Scan sensor (figure 6.1a) until push-off of the contralateral foot (figure 6.1b) when there was a significant forward shift in load.

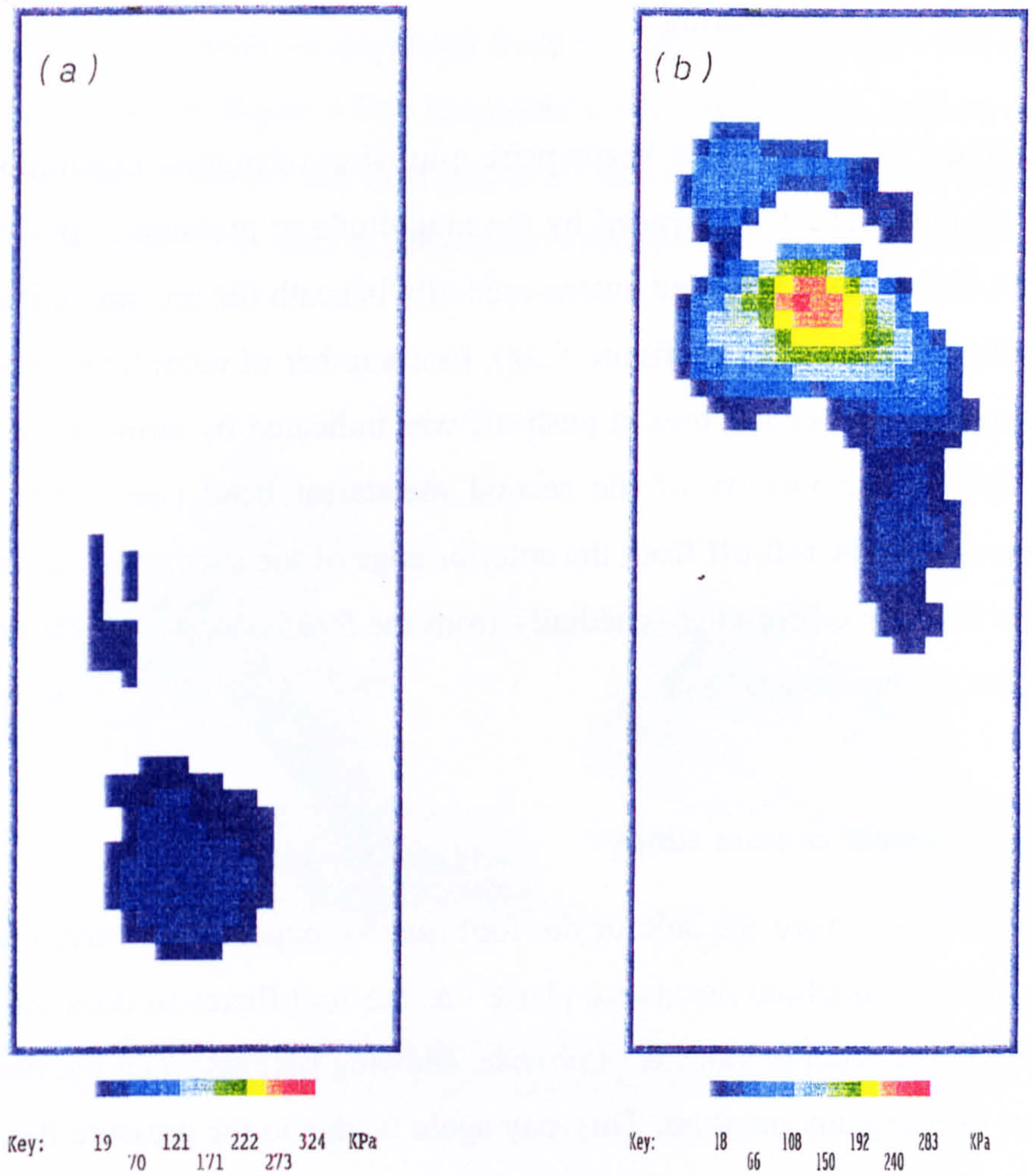

Figure 6.1. Onset of pressure beneath the centre of the heel at heel-contact during shod walking (a), which is coincident with push-off of the contralateral foot (b).

Pressure beneath the midfoot throughout the stance phase was, in most cases, only a fraction of the pressures beneath the heel and forefoot (see figure 5.3). As a result, the locus of the CoP beneath the midfoot was largely influenced by the onset of pressure beneath the forefoot. Progression of the locus just lateral to the midline was associated with a central onset of forefoot pressure; while progression of the locus along the midline was associated with a medial onset of forefoot pressure (see figure 5.2a). The progression of the locus beneath the midfoot close to the midline in shod walking suggests reduced inversion of the foot, which may be the norm when Gibson style shoes are worn as compared to barefoot walking. Hence in Gibson style shoes there appears to be a rapid full contact between the forepart of the shoe outsole and ground after full heel-contact and not a gradual lateral to medial loading of the forefoot as seen in barefoot walking.

The second and third metatarsal heads were joint sites of highest maximum peak pressure beneath the forefoot. Influenced by the magnitude of pressure in this region, the locus of the CoP in-shoe travelled antero-medially beneath the second and/or third metatarsal heads in most cases (see figure 5.2a). In a number of recordings, elevation of the foot from the forefoot and toes at push-off was indicated by termination of the locus of the CoP in the vicinity of the second metatarsal head (see figure 5.2a). However, in most records, roll-off from the anterior edge of the shoe was indicated by the locus of the CoP travelling antero-medially from the forefoot to terminate beneath the lateral aspect of the hallux.

6.3.2 foot-to-shoe insole contact times

When lace-up shoes are worn the sole of the foot may be expected to exert pressure over the shoe insole throughout the stance phase - as the foot flexes so does the shoe. However, the pressure records indicate otherwise, showing periods when the heel and forefoot are not exerting any pressure. This may again be due to the pressure threshold of the sensor that was used for measurements.

The durations of local plantar contact with the shoe insole that were measured were longer than the contact times measured in barefoot walking by other workers. However, as a result of the threshold of the pressure sensor used, the onset and cessation of pressure at local plantar sites was probably earlier and later into the stance phase, respectively, than that measured. In barefoot studies of healthy subjects workers have reported the heel, forefoot and hallux to be in contact with the ground for between 45-65%, 63-87% and 56-60% of the stance phase, respectively (**Scranton and McMaster, 1976; Hutton and Dhanendran, 1981; Soames, 1985**). In the present study the average contact times for these areas was 64.4%, 87.5% and 73.3%, respectively (see figure 5.4a). The hypothesised interaction between the foot and lace-up style shoe dictates that local plantar contact with the shoe insole is maintained for longer than these times. In the case of the heel, although there is a tendency for the heel pad to lift off the shoe insole between heel-lift and push-off, this will be prevented by pressure on the instep from the upper, and pressure and shear on the heel from the counter (figure 6.2a). Increased toe-spring will also reduce the tendency for the heel pad to lift off the shoe insole at push-off by minimising flexion of the shoe across the tread line with a subsequent reduction in pressure on the instep and heel (figure 6.2b).

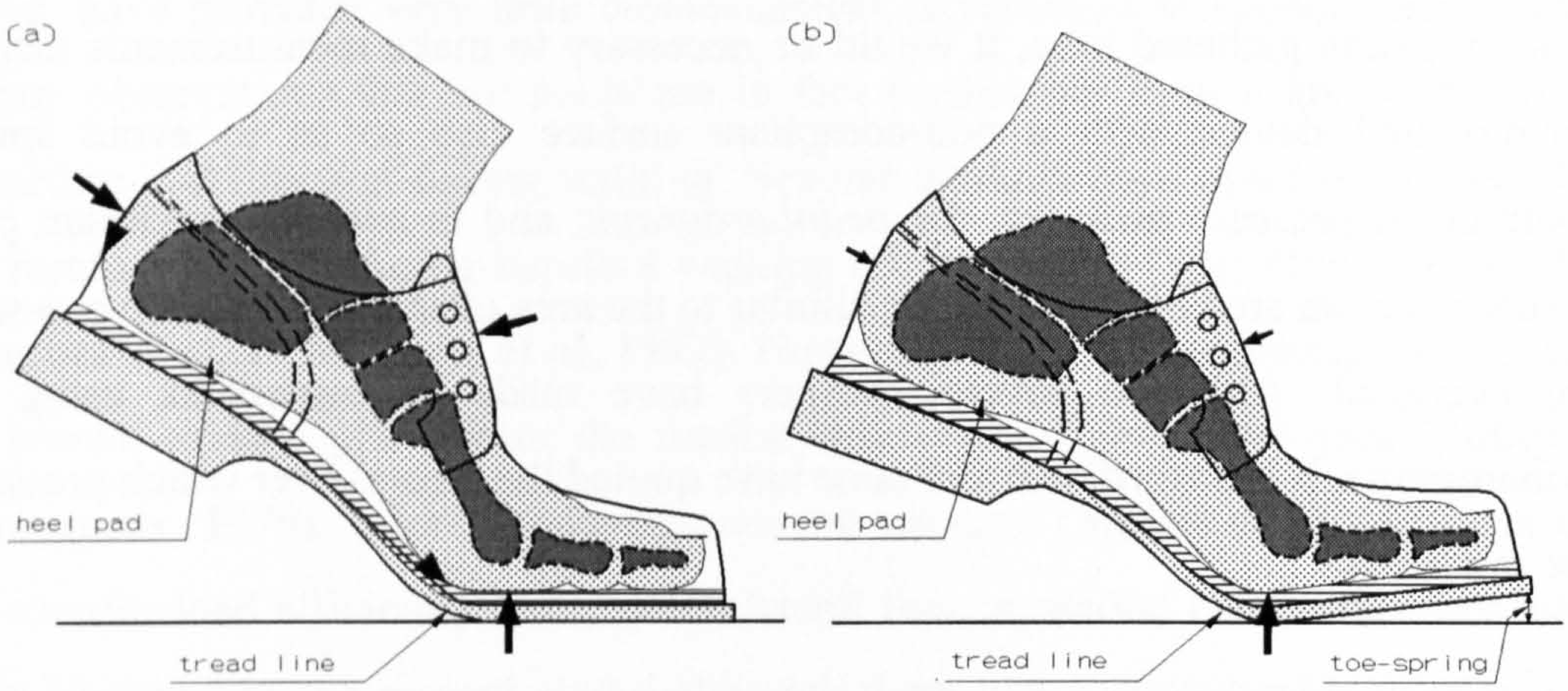

Figure 6.2. Tendency for the heel pad to lift off the shoe insole at push-off: forces exerted by the foot on the shoe (a), which are reduced by increased toe-spring (b).

The extended loading of the forefoot and toes when shoes are worn is due to the forepart of the shoe outsole making contact with the ground much earlier into the stance phase than in barefoot walking. Increasing heel height has been found to result in a progressively earlier contact between the forepart of the shoe outsole and ground after heel-contact (**Snow et al, 1992**), probably as a result of the foot remaining plantarflexed throughout the stance phase in combination with reduced knee extension, hip flexion and step length.

6.3.3 local maximum peak pressures

Although there have been many studies documenting local maximum peak pressures beneath the asymptomatic foot during barefoot walking (see table 2.3), none of these can be used to make a valid comparison with the local maximum peak pressures measured in the present in-shoe study. This is because the conditions under which barefoot measurements were made and the analyses that were performed on the data gathered are not comparable to those employed in this study. In the present study, in-shoe plantar pressures were measured beneath the forefoot and heel while these parts of the foot were supported by a non-compliant surface, ie. high-density Plastazote (see figure 4.4a). In order to provide barefoot plantar pressure data that *is* comparable to the in-shoe data gathered here, it would be necessary to make measurements using a floor-mounted device with a non-compliant surface also so as to avoid spatial smoothing of pressure peak, eg. the *pedobarograph*; and to average maximum peak pressures over an area of 2.25 cm^{-2}, ie. similar to the area over which in-shoe pressures were averaged. Although several workers have made measurements using the pedobarograph, it is unfortunate that none have quoted the area(s) over which pressures were averaged.

Two workers have measured barefoot plantar pressures using discrete transducers which had similar areas to the area over which in-shoe pressures were averaged (**Pollard, 1984,** 1.76 cm^{-2}; **Soames, 1985,** 1.69 cm^{-2}). However, when the methods employed by these workers, with regard to the conditions under which measurements were made, are taken into account it is no longer possible to make a valid comparison between their data (see table 2.3) and the data from the present study. In both of these earlier studies, measurements were made with transducers taped to the sole of the foot beneath anatomical sites that were located by palpation. It was recently shown that this method of locating discrete transducers beneath the foot mislocates areas of peak pressure beneath the metatarsal heads, which move anteriorly during walking (**Lord et al, 1992**). It is also apparent that the surface across which the subjects walked was not standardised: subjects walked across a 1.5 mm thick rubber mat (**Pollard**) or across a wooden floor (**Soames**). In the former study, pressures would have been reduced due to spatial smoothing when walking on the compliant rubber surface; whereas in the latter study, pressures would have been increased due to concentration of pressure at the sites of the transducers when walking on the hard wooden floor. It is possible to conclude that the pressures measured by **Pollard** and **Soames** are underestimates and overestimates, respectively, of the maximum peak pressures that would be measured over similar areas by the pedobarograph - the device thought to be ideal for barefoot measurements. However, all this aside, local maximum peak pressures are thought to be lower beneath the shod as opposed to barefoot due to the cushioning typically provided by the shoe insole.

Most workers who have conducted plantar pressure studies, whether barefoot or shod, have provided very little biomechanical information to support their findings. Many observations that are made are in fact predictable from a knowledge of foot structure and function during walking. Several workers have observed a medial bias in forefoot loading during barefoot walking (**Betts et al, 1980c**; **Grieve et al, 1984**; **Hughes et al, 1990**; **Snow et al, 1992**). However, this is not surprising since push-off is known to take place from the medial side of the forefoot (**Bojsen-Moller and Lamoreux, 1979**), where the metatarsals are stronger (**Stokes et al, 1979**), in order to transfer load efficiently to the contralateral foot. A medial bias in forefoot loading was observed in the present study, although there was a considerable overlap in the ranges of maximum peak pressures measured beneath the first, second and third metatarsal heads. This overlap is thought to be due to spatial smoothing of local pressure peaks, brought about by the use of an insole sensor rather than individual discrete transducers.

The angle of gait and forward projections of the metatarsal heads may influence the order of forefoot loading. In the present study, local maximum peak pressures occurred in a progressive order beneath the forefoot of the subjects reflecting a gradual transfer of load from lateral to medial (see figure 5.4a). In comparison, **Dhanendran (1979)** noted maximum peak pressures to occur in a different order beneath the barefoot: mth(3-5)[38]-mth1-mth2. From this order it appears that the subjects in this latter study had a very narrow angle of gait, maybe even with the imaginary midline

[38] The 3rd, 4th and 5th metatarsal heads were grouped together for analysis.

of the foot parallel to the direction of progression. With a narrow angle of gait, the metatarsal head projected the furthest forward is expected to be the last at which the peak pressure reaches a maximum as the foot inclines relative to the support surface in readiness for push-off. The second metatarsal head is commonly projected further forward than the other four **(Viladot, 1991)**, hence the timing order noted by **Dhanendran** possibly indicates the majority of the subjects in his study group had an index minus type forefoot.

6.4 COMPARISON OF THE NORMATIVE IN-SHOE PLANTAR PRESSURE DATA GATHERED DURING THE CLINICAL TRIAL AND DATA FROM OTHER IN-SHOE PLANTAR PRESSURE STUDIES

The basis for this comparison was to validate the plantar pressure data gathered during the clinical trial. However, from the literature review of Chapter 2 it is apparent that there is a lack of comprehensive in-shoe plantar pressure data despite the fact that shoes are routinely worn by the majority of people in the western world. Quantities such as the locus of the CoP in-shoe, local pressure-time integrals and local foot-to-shoe insole contact times have not been studied in great detail.

6.4.1 foot-to-shoe insole contact times

One other worker has studied the durations of local plantar contact with the shoe insole during the stance phase. **Soames (1985)** made measurements using discrete transducers which had a pressure threshold similar to the sensor used in the present study - between 20 kPa and 30 kPa. It has already been remarked that the sole of the foot is thought to exert pressure over the insole of the shoe *throughout* the stance phase. However, when plantar pressures are measured using sensors/transducers that have a high pressure threshold the durations of local plantar contact with the shoe insole are underestimated. Because of this, it would be more precise to refer to the *durations of local plantar contact* as the durations when local plantar pressure exceed the threshold level of the measuring device.

The measurements made in the present study are comparable with the findings of **Soames (1985)** (table 6.1). In both studies, the pressures measured were in excess of the threshold levels of the measuring devices for a longer period beneath the lateral side of the forefoot compared to the medial side. In the present study this was due to an earlier onset of pressure beneath the lateral side of the forefoot (see figure 5.4a), which suggests that the foot was in an inverted position at heel-contact.

Table 6.1. Durations of local plantar contact with the shoe insole, expressed as a percentage of the stance phase, in the subject group.

Study	Durations of local plantar contact with the shoe insole (expressed as a percentage of the stance phase)						
	hallux	mth1	mth2	mth3	mth4	mth5	heel
present study	73.3	78	81.8	83.9	83.5	-	64.2
Soames (1985) (average of men and women, estimated from graph)	80.6	82.6	76.2	81.3	84.0	85.9	67.1

6.4.2 local maximum peak pressures

At first glance the local maximum peak pressures measured in the present study do not compare well with the findings of other workers (compare tables 5.10 and 2.4). This is thought to be because of the various sizes of transducer/analysis areas over which pressures were averaged and the different study methods which were employed. **Rose et al (1992)** averaged pressures over very large areas beneath the forefoot, grouping together the first and second, and the third, fourth and fifth metatarsal heads. In the studies of **Pollard (1984)** and **Soames (1985)**, the plantar surface was palpated to locate anatomical landmarks and discrete transducers were subsequently taped directly to the foot. The effect that this method of transducer location and attachment has on the magnitudes of local maximum peak pressure that are recorded has already been discussed (see section 6.3.3). The study of **Zhu et al (1991)** was similar in design to the present study although measurements were made with discrete transducers. The transducers were embedded into an insole, which avoids the local concentration of pressure that would otherwise occur; and were also accurately located beneath areas of maximum peak pressure as determined by an ink-printing method. The area of the transducer used by these workers was a factor of 2.4 smaller than the area over which local pressures were averaged in this study, while the maximum peak pressures they measured were on average a factor of 2.8 larger. The linear relationship between local maximum peak pressures and transducer area that is implied by this comparison is, however, not intended, particularly in the case of patients with diabetes who have healed neuropathic ulcers. When local peaks of pressure generated beneath the sites of healed ulcers are seen in 3-D they often appear as spikes, which does not warrant a linear relationship between these two quantities (**Lord, 1993**, figures 6.3 and 6.4).

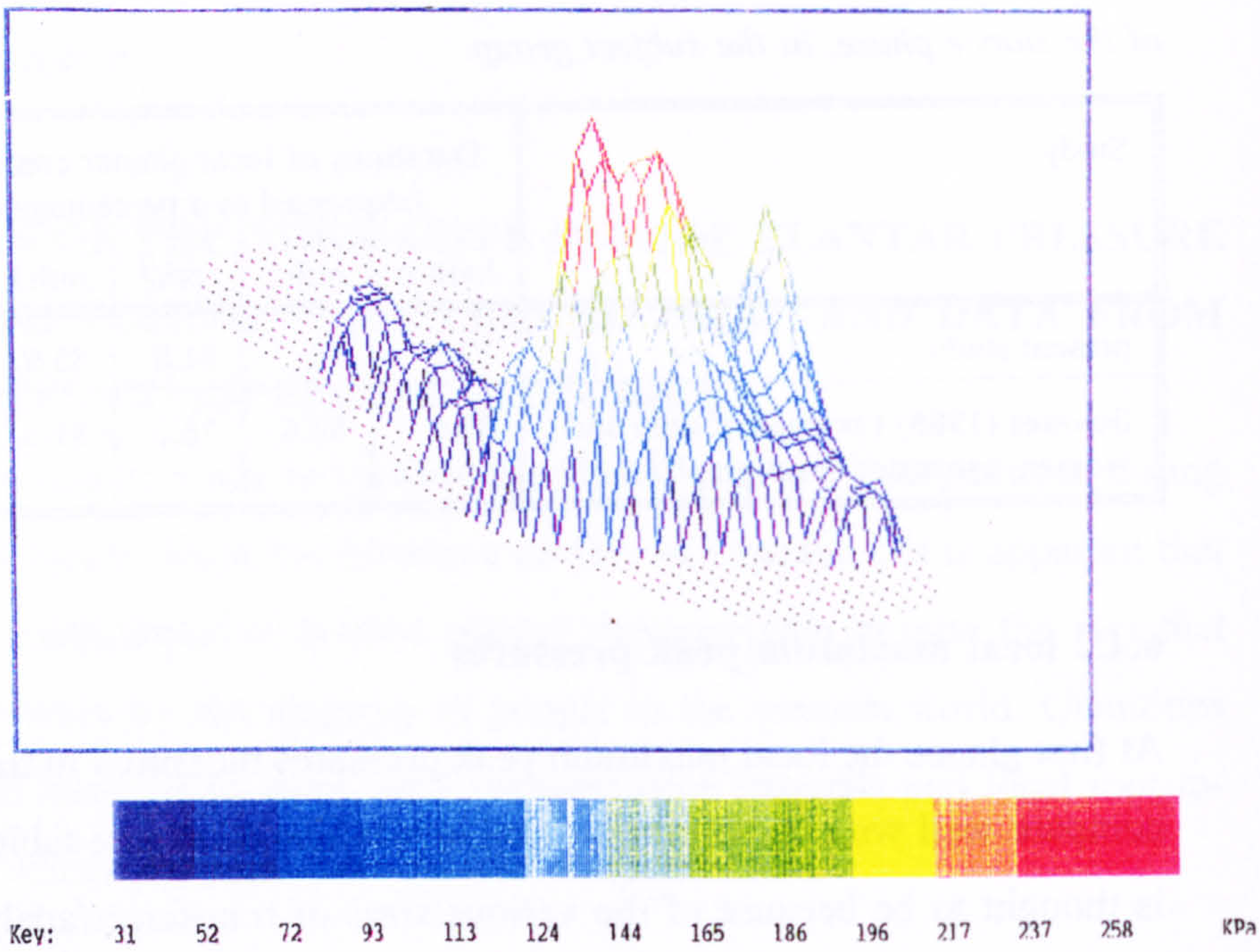

Figure 6.3. 3-D image of the pressure generated beneath the forefoot at push-off, for a patient with diabetes who has a healed neuropathic ulcer beneath the second metatarsal head. The frame shown here in 3-D is the same frame shown in 2-D in figure 2.16a.

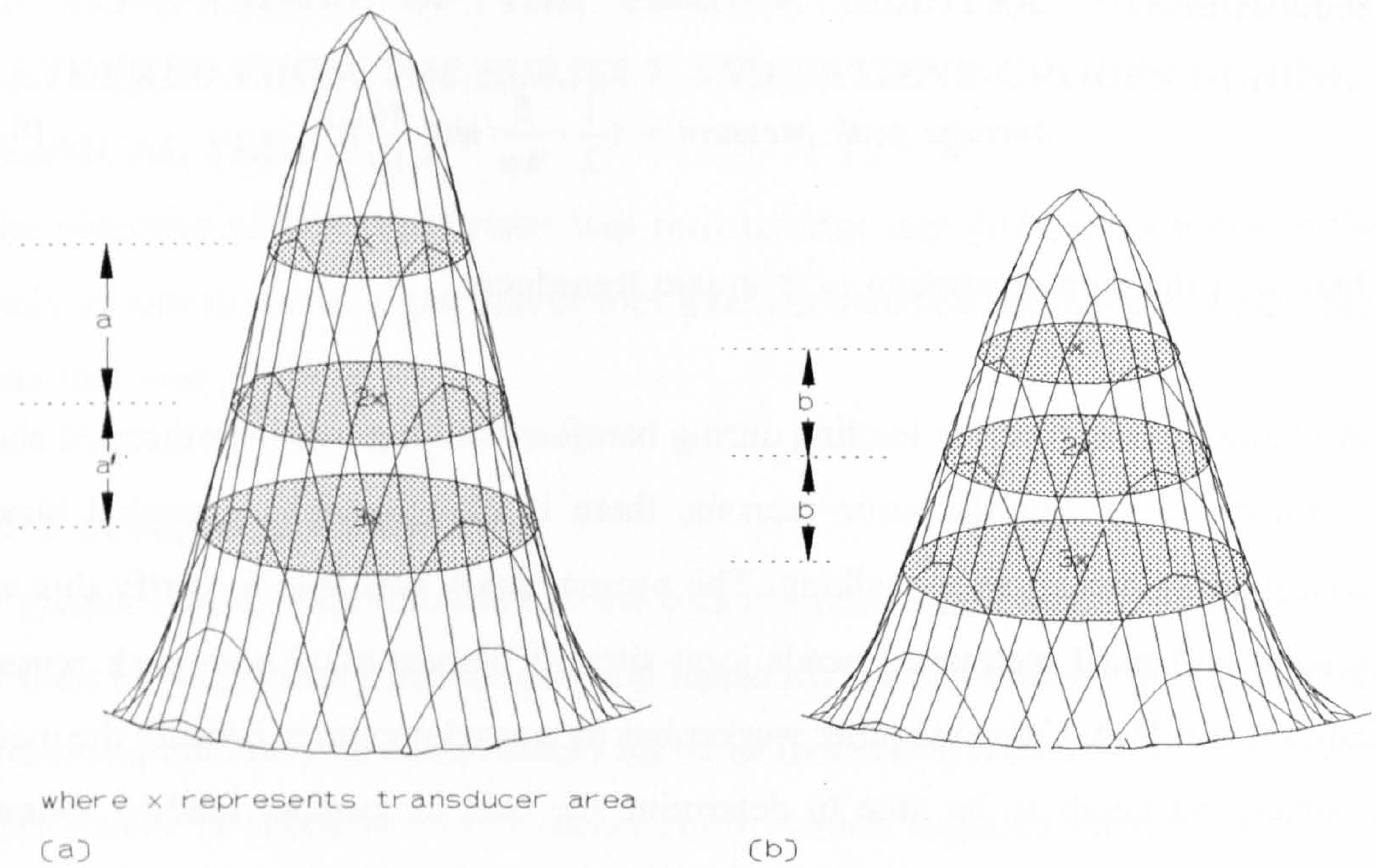

Figure 6.4. 3-D profiles of the pressure peaks beneath the metatarsal heads: (a) typical profile (b) profile assuming linearity between transducer area and peak pressures.

The relationship hypothesised, for an example peak, between average peak pressure and transducer area was determined by **Lord (1993)** in a theoretical analysis. This analysis was based on the assumption that the pressure profile was symmetric in all directions, similar to that seen in figure 6.4b, and expressed by equation 6.1.

$$p = 0.5(1+\sin(\frac{2\pi}{20}x-\frac{\pi}{2})).\ 0.5(1+\sin(\frac{2\pi}{20}y-\frac{\pi}{2}) \qquad \textbf{(6.1)}$$

With the assumption that the transducer was square, with dimensions 2a x 2a, the pressure averaged over a square area centred on the peak is given by

$$\frac{1}{4}a^2 \int_{10-a}^{10+a} \int_{10-a}^{10+a} p\ dx\ dy \qquad \textbf{(6.2)}$$

which expands to

$$[\frac{1}{4}a^2\ (x+\frac{20}{2\pi}\cos(\frac{2\pi}{20}x-\frac{\pi}{2}))\ ]_{10-a}^{10+a}\ \ (y+\frac{20}{2\pi}\cos(\frac{2\pi}{20}) \qquad \textbf{(6.3)}$$

and reduces to

$$\textit{average peak pressure} = \left(\frac{1}{2}+\frac{5}{\pi a}\sin\left(\frac{\pi a}{10}\right)\right)^2 \qquad (6.4)$$

where a is the side dimension of a square transducer.

A medial bias in forefoot loading during barefoot walking was hypothesised above (see section 6.3.3). For the same reasons, there is thought to be a medial bias in forefoot loading during shod walking. The present study was able to verify this with the second and third metatarsal heads joint sites of highest maximum peak pressure beneath the forefoot. Only one other worker has measured pressures beneath the medial four metatarsal heads to be able to determine the bias in forefoot loading. **Soames (1985)** found a more pronounced medial bias in forefoot loading than the present study, with the first metatarsal head the site of highest maximum peak pressure beneath the forefoot of both men and women (see table 2.4). **Soames** had allowed his subjects to wear their own shoes, which were not described. A medial bias in forefoot loading is consistent with the women wearing high-heeled shoes - **Snow et al (1992)** found a similar bias in forefoot loading to **Soames** when women wore shoes with a 8.26 cm heel. However, apart from a very wide angle of gait, the reason for the pronounced medial bias in forefoot loading in the male group is difficult to ascertain.

A discrete area of high pressure associated with the fifth metatarsal head was rarely seen in the pressure records. In a few records this was due to the generally low pressures encountered in this area. However, in a majority of records the area of high pressure associated with the fourth metatarsal head was close to the lateral edge, implying the fifth metatarsal head was overhanging the edge of the shoe outsole. It is thought that the outsole of the trial stock shoes was too narrow to accommodate the splaying of the metatarsals upon forefoot loading so that the fifth metatarsal head ended up resting on the inside of the upper. This is in fact a normal condition with fashion shoes.

6.5 COMPARISON OF THE IN-SHOE PLANTAR PRESSURE DATA GATHERED FROM THE SUBJECT AND PATIENT GROUPS DURING THE CLINICAL TRIAL

The objective of this comparison was to determine any differences between the two study groups in terms of gait and/or foot function based on the in-shoe plantar pressure data that was gathered.

6.5.1 timing of gait phases

A greater stance phase duration in the patient group was expected as a consequence of their plantar sensory deficit: postural instability, which may occur due to abnormal proprioception, may be compensated for by an increase in foot-to-ground contact time. This theory is supported by a recent study, which suggests patients with diabetic neuropathy have poor body sway control compared to healthy subjects when performing the *Romberg* manoeuvre, ie. raising the arms from the sides of the body to a horizontal position, while standing (**Boulton et al, 1994**). The Romberg manoeuvre is intended to induce a disturbance to balance, perhaps similar to that which normally occurs with each step during walking. In the present study, there was found to be no significant difference in the duration of the stance phase between the two groups. Instead, an increase in the duration of the stance phase may be a consequence of ageing: as an individual gets older their speed of walking may become slower, a view supported by **Gabell and Nayak (1984)**. Age was found to be positively correlated with the duration of the stance phase in both groups studied (subjects: r=0.835; patients: r=0.652, see figure 5.1), but this was not significant in the patient group - the patient group may have been too small (n=7) or ages not spread over sufficient range (48 to 78 years) to obtain a significant result.

6.5.2 locus of the centre of pressure

In most records the locus of the CoP originated beneath the central or anterior aspect of the heel instead of the postero-lateral aspect as expected, probably as a result of the pressure threshold of the insole sensor used (see section 6.3.1). In three patient records the locus originated beneath the posterior aspect of the heel (see figure 5.2b). In these cases, pressure beneath this area of the heel probably exceeded the threshold level of the sensor *before* the heel of the shoe made full contact with the ground. Although this implies heavy heel-contact, which may be related to diminished plantar sensation and proprioception, one patient in whose record this was observed had normal plantar sensation.

In one patient the locus of the CoP travelled medial to the midline of both feet as a result of pressure increasing more rapidly beneath the medial rather than the lateral aspect of the forefoot (see table 5.9). Also, in most records the expected lateral to medial progression of the locus of the CoP across the forefoot was not seen. Both of these observations suggest a reduced capacity for the foot to invert at the subtalar joint in shod as compared to barefoot walking. In fact the amount of subtalar inversion that *is* allowed in shod walking may be largely dependent on the style of the shoe.

Patients had a tendency to push-off from the forefoot, as indicated by the terminal point of the locus of the CoP; and rarely rolled forward onto their toes to push-off directly from the hallux, as did most subjects. Most patients in the study had clawed or retracted toes - deformities which are known to result in reduced toe loading and increased forefoot loading (**Ctercteko et al, 1981**). Just as likely, the tendency for the patients to use the forefoot at push-off may be related to the shoes that were worn. Most of the patients in the present study normally wear rocker-inserts, which reduce flexion across the treadline and the subsequent roll onto the toes at push-off. Although these inserts were removed during pressure measurements, the matching rigidity and contours of the shoe outsole would produce the same effect as the inserts.

6.5.3 foot-to-shoe insole contact times

Between the two groups studied there was no significant difference in the times of onset, peak and cessation of pressure at local plantar sites with one exception: the onset of pressure beneath the hallux occurred significantly later into the stance phase in the patient group (see figure 5.4). Although a number of patients had clawed or retracted toes, which may have reduced the weight-bearing role of the hallux - a view supported by other studies (**Stokes et al, 1975; Ctercteko et al, 1981**) - the increased toe-spring of the patients' shoes may also have had the same effect.

Many workers consider moderate stresses repetitively applied for long periods of walking to be a significant risk factor in plantar ulceration (**Oakley et al, 1956; Stokes et al, 1975; Boulton et al, 1983; Delbridge et al, 1985, Lemkes et al, 1994**). It would follow from this hypothesis that plantar ulceration may be more likely to occur at sites of extended loading during the stance phase. It was anticipated that the forefoot valgus condition noted in a majority of patients would be associated with extended loading of the medial rather than the lateral side of the forefoot; and with a propensity for ulceration beneath the medial side of the forefoot. Out of a total of eight healed ulcers located beneath one of the medial three metatarsal heads[39] only two were associated with a forefoot valgus condition *and* with extended loading beneath the medial rather than the lateral side of the forefoot (table 6.2). However, it is interesting to note that a forefoot valgus condition was assessed in seven out of eight feet in which ulceration had occurred beneath one of the medial three metatarsal heads.

Table 6.2. Durations of local plantar contact with the shoe insole, expressed as a percentage of the stance phase, in the patient group[a] (bold type indicates the site of highest maximum peak pressure beneath the forefoot; the use of underline indicates sites of healed ulcers)[39].

Patient-foot	Durations of local plantar contact with the shoe insole (expressed as a percentage of the stance phase)						Forefoot condition
	hallux	mth1	mth2	mth3	mth4	heel	
1-R	41.9	77.1	75.1	71.3	65.4	74.1	plantarflexed 1st metatarsal type forefoot valgus
1-L	41.2	**43.4**	44.9	50.0	58.7	66.9	total-type forefoot valgus
2-R	50.4	49.4	**56.0**	79.9	81.0	57.5	forefoot varus
3-L	77.3	**72.0**	82.1	83.0	77.9	66.0	plantarflexed 1st metatarsal type forefoot valgus
4-L	84.1	**76.4**	82.4	87.9	86.0	70.0	total-type forefoot valgus
5-L	54.3	91.4	**93.9**	**93.9**	86.8	59.4	plantarflexed 1st metatarsal type forefoot valgus
7-R	67.8	75.4	**80.1**	83.1	72.9	46.6	total-type forefoot valgus
7-L	43.4	83.1	**83.7**	89.8	88.6	63.3	total-type forefoot valgus

[a]from the mean data presented in Appendix E.

[39] Patient number 6 is not included in this analysis as creasing of the F-Scan sensor during pressure measurements made the data recorded unreliable. The ulcer beneath the forefoot of patient number 5 was located in the region of the 2nd/3rd metatarsal heads.

6.5.4 local maximum peak pressures

In the present study, maximum peak pressures beneath the hallux, heel and the first, second, third and fourth metatarsal heads were found to be similar between the two groups. This result is contradictory to many barefoot studies, and the only other shod study of similar groups (**Pollard, 1984**), which have found maximum peak pressures to be higher in the patient group. The methodology used by **Pollard** was similar to that used here with regards to the shoewear worn by individuals during measurements: the subjects wore one style of shoe, whereas the patients wore their own. However, **Pollard** made measurements with discrete transducers, in contrast to a pressure sensitive insole used here. Measurements were made with the insole sensor placed on top of an inlay, which provided the foot with a large area of support in the regions of the heel and forefoot, and which may have resulted in considerable spatial smoothing of pressure peaks. In addition, the large area over which pressures were averaged may have resulted in considerable spatial averaging of localised peaks of pressure, especially beneath the feet of the patients.

Group averages of the local maximum peak pressures indicate a bias towards medial loading of the forefoot in the patients similar to the subjects as expected (see section 6.3.3). Most healed plantar metatarsal ulcers that were noted in the patients were beneath one of the three medial metatarsal heads and were coincident with the site of highest maximum peak pressure beneath the forefoot (see table 6.2). A particularly important point arises from these observations since it is apparent that shoes which reduce the tendency towards medial loading of the forefoot may in fact reduce the risk of plantar ulceration in patients with diabetes.

A plantarflexed first metatarsal condition was assessed in four out of five feet in which ulceration had occurred beneath the first metatarsal head; while a plantarflexed fifth metatarsal condition was assessed in the one foot in which ulceration had occurred beneath the fifth metatarsal head (see table 5.25). Both of these conditions are associated with a greater range of metatarsal movement in plantarflexion than in dorsiflexion, which may be a significant risk factor in ulceration beneath these metatarsal heads. It can be hypothesised that a plantarflexed first or fifth metatarsal may result in the generation of abnormally high plantar pressures beneath these respective metatarsal heads because these metatarsals have a limited capacity for upward movement upon loading. To support this view, three out of the five healed ulcers noted beneath the first metatarsal head of the patients in the present study were coincident with the site of highest maximum peak pressure beneath the forefoot and were also associated with a plantarflexed first metatarsal (table 6.2). It is apparent that the assessment of first and fifth metatarsal mobility may be a good risk indicator of plantar ulceration beneath these respective metatarsal heads.

6.6 SUMMARY

This Chapter has provided a comparison of: (i) barefoot and shod foot function and (ii) shod foot function in asymptomatic subjects and patients with diabetes, based on the measurement of plantar pressure. It has been highlighted that workers who have recorded plantar pressures have provided very little biomechanical information to support their findings. This is considered somewhat unfortunate since there is little understanding of how plantar pressures are generated and more importantly how particular styles of shoewear may be better than others in reducing the risk of plantar pathologies.

To conclude this Chapter, the findings of the present study will be briefly summarised.

Contrary to many barefoot and in-shoe plantar pressure studies, similar magnitudes of maximum peak pressure were measured in the two groups of the present study.

The pressure data recorded indicated a medial bias in forefoot loading in the patient group and more central loading of the forefoot in the subject group - a finding which corresponded with a high incidence of healed plantar ulcers beneath the first and second metatarsal heads of the patients.

The most common path taken by the locus of the CoP in-shoe beneath the forefoot suggested two interesting differences between barefoot and shod foot function. Firstly, the path noted beneath the central aspect of the forefoot indicated a limited capacity for the foot to invert at the subtalar joint at heel-contact in shod walking. Secondly, the terminal point of the locus indicated that patients push-off from the forefoot and the subjects push-off from the hallux.

CHAPTER 7
DISCUSSION:
IN-SHOE PLANTAR SHEAR STRESS MEASUREMENT

7.1 INTRODUCTION

The ground reaction force, which is exerted by the body through the foot during walking, consists of vertical and horizontal components. The horizontal components in the antero-posterior and medio-lateral directions, which are generated during shod walking and which may be recorded by most modern forceplates, provide evidence for friction and hence shearing between the shoe outsole and ground. This shear may be transmitted to the foot as either shear or pressure on the plantar and dorsal aspects.

Until very recently there was a lack of suitable technology for the development of devices for measuring shear. Consequently, there have been very few studies of in-shoe plantar shear stresses. The most comprehensive of these studies to date has been that of **Pollard (1984)** who made measurements with discrete transducers taped to the sole of the foot. The interposition of a rigid transducer between the foot and shoe insole undoubtedly has a perturbing effect on the magnitudes of the local plantar shear stresses that are generated. However, although this was acknowledged by **Pollard** it is a methodology that has since been used by others (**Tappin and Robertson, 1991, Laing et al, 1992, Lebar et al, 1993**). A second source of inaccuracy in measuring maximum peak shear stresses is also apparent in all previous studies and arises from the static calibration of transducers which are then used to make dynamic measurements.

For the above reasons there is presently no reliable data as to the patterns and magnitudes of the local plantar shear stresses that are generated during shod walking. This is a possible hinderance to many workers who seek to determine the role of shear in causing tissue damage and in particular plantar ulceration of the insensitive diabetic foot. In the present study, the method of flush-mounting discrete shear transducers into an insole was developed in order to minimise perturbation effects; and the transducers were quasi-statically calibrated prior to making dynamic measurements.

With a view to understanding the role of shear in causing ulceration of the insensitive diabetic foot, local plantar shear stresses were measured at the foot/shoe interface of two groups: asymptomatic subjects who wore standard extra-depth orthopaedic shoes and patients with diabetes who wore their own bespoke shoes. This Chapter discusses the results obtained, first with reference to the methods and measuring equipment employed; then with reference to biomechanical interpretation and signal analysis.

7.2 APPRAISAL OF THE STUDY GROUPS, METHODS AND EQUIPMENT

Quite apart from the obvious individual factors such as deformation of soft tissue and sliding of the foot in the shoe during walking, there are other *study-design* related factors which are important to consider for their possible role in influencing the patterns and/or magnitudes of local plantar shear that were recorded. Hence as a basis to a discussion of the shear records presented in the Appendices, consideration will be given to the different characteristics of the two groups studied and the limitations of the methods and measuring equipment employed.

7.2.1 study groups and methods

Recall from Chapter 6, section 6.2.1, the discussion of the difficulties encountered in recruiting the patients to the clinical trial which led to the study of two groups with different characteristics. Small groups were studied which were not matched in numbers or for age, body mass or sex: the patients were older and heavier than the subjects and there was a predominance of men in both groups. A review of the literature revealed several studies which have found no correlation of local maximum peak pressures with either age or body mass; and no significant difference in the local maximum peak pressures beneath the feet of men and women (see Chapter 2, section 2.5.6). In contrast to those for plantar pressures, the relationships between local plantar shear stresses and either age or body mass have not been studied. Unfortunately it was not possible to investigate these relationships in the present study because of the small numbers in the groups.

Local plantar shear stresses were not recorded beneath the medial four metatarsal heads simultaneously. **Pollard (1984)** had been able to do this by taping transducers to the foot in close proximity. However, for reasons discussed in detail in Chapter 4, current measurements were made with transducers flush - mounted into an inlay. Although the diameter of the transducers which were used in the present study were identical to those used by **Pollard**, the diameter of the inlay mounting did not allow the transducers to be placed in close proximity. As a result shear stresses were recorded beneath the medial four metatarsal heads over two successive walks with duplicate measurements made beneath the heel. A simple visual inspection of the shear records revealed similar patterns of shear and magnitudes of maximum peak shear stress were generated beneath the heel during the two successive walks which lends weight to these two walks being similar (see Appendices F and G).

7.2.2 equipment

The integrity of the signal leads attached to the shear transducers was maintained throughout the clinical trial. In the study of **Pollard (1984)** lead breakages at the point of entry into the transducer were a common occurrence as a result of the leads being placed in tension during weight-bearing. In the present study, the protection offered to the transducers by the inlay mounting was sufficient to avoid this mode of damage. One transducer was, however, damaged through separation at the rubber/acetal copolymer interface (the x-axis of transducer number 1, see tables 3.2 and 3.3). This transducer had been subjected to fewer than 100,000 gait cycles (the design lifetime (**Williams, 1993**)) and is thought to have sheared apart in this way due to ageing of the cyanoacrylate glue bonding the two materials.

During data collection, signal-conditioning was performed via pre-amplifier and filter units. Pre-amplification electronics were contained in a box which was worn as a backpack and which was connected via a trailing ribbon cable to the unit providing low pass filtering. Although it was not ideal for individuals to carry a backpack, it was necessary to amplify the transducer signals above potential noise levels *before* conveying them via the ribbon cable to the filter unit. Individuals did not complain of any discomfort due to carrying the backpack; and the force required to drag the ribbon cable connecting the pre-amplifier unit to the filter unit (approximately 2.5 N) did not appear to pose a problem either (see figure 4.9).

The signal-conditioning electronics were originally designed to interface ten tri-axial transducers (for measuring two orthogonal directions of shear, and pressure) and to provide a filter cut-off frequency of 250 Hz (**Williams, 1993**). For the present study the signal-conditioning electronics were modified in both these respects. In order to reduce the weight of the backpack four transducer interfacing circuits were removed. Although data collection was enabled from a maximum number of three shear transducers at any one time during this study, six transducer interfacing circuits were necessary for a parallel study being conducted by a co-worker measuring tri-axial stresses at the stump-socket interface of patients with lower limb amputations. The filter electronics were redesigned with a lower cut-off frequency of 100 Hz after considering the maximum frequency content of the shear signal that would be of interest; and after dynamic tests indicated possible resonance of a shear axis of the transducer at approximately 100 Hz (*op cit*).

Several improvements in the design of the shear transducer and the interfacing electronics are envisaged. Consequently, the equipment used in the present study is thought unsuitable for further measurements. With regard to the design of the shear transducer, there are two limitations. Firstly, as a result of the physical configuration of the signal leads that are attached, the transducers cannot be placed in close proximity; secondly, their thickness currently hinders the development of a thin inlay - measurements can only be made inside extra-depth shoes at present. Improving the design of this transducer (based on the magneto-resistive principle) is not thought to be potentially rewarding. Instead it is thought more practical to seek another principle of transduction for the next generation of shear transducer. For example, it is currently possible to manufacture a smaller transducer utilising piezoelectric crystals, although at a high cost. With regard to the interfacing electronics, the size and weight of the pre-amplifier backpack could be made more portable with the use of existing electronic hardware; and the trailing signal wires could be abandoned in favour of telemetry. In this way it would be possible to make measurements over longer walks as well as during vigorous activities such as running and jumping.

7.3 SHEAR RECORDS

7.3.1 coding system for the shear records

For brevity, the shear records in the Appendices are referred to in the ensuing discussions using a simple five character code:

- *S* or *P* refers to a Subject or Patient shear record, respectively;
- a number refers to the particular subject or patient (see tables 5.1 and 5.2, respectively);
- *l* or *t* refers to a record of longitudinal or transverse shear, respectively;
- *R* or *L* refers to the Right or Left foot, respectively;

and

- *s* or *b* refers to the condition under which the recording was made, either with the individual wearing or not wearing nylon hold-ups, respectively, while in the extra-depth stock orthopaedic shoes ie. either wearing socks or barefoot.

As an example, *S9tRs* refers to the shear record of Subject number 9, for transverse shear stresses recorded beneath the Right foot while wearing nylon hold-ups (socks). In cases where it is necessary to refer to both the longitudinal and transverse shear records of an individual the letters *l* and *t* appear in brackets, eg. *S9(lt)Rs*.

Repeat shear tests were carried out on one subject and one patient three months apart to assess repeatability. The shear records of these tests are referred to by the addition of the letter *R* to the end of the code described above, eg. *S9lRs-R*.

7.3.2 choice of features to analyse in the shear records

In previous studies of local plantar shear, workers have either concerned themselves with quantifying the time to maximum peak longitudinal shear as a percentage of the stance phase **(Tappin and Robertson, 1991)** or with quantifying "peak-to-peak" longitudinal and transverse shear stresses **(Pollard, 1984)**. The time to maximum peak shear stress can be a misleading quantity because the peak beneath, say the first metatarsal head, may occur at heel-contact, push-off or at a time in-between due to inter-subject variability in the pattern of the waveforms generated. Presenting peak-to-peak values of shear stress can also be misleading: whereas peak-to-peak values may be similar at two sites there may be different magnitudes of say, anterior shear (eg, *S8lRb* - see waveforms for shear recorded beneath the first and second metatarsal heads). Although it is interesting to know the "peak-to-peak" longitudinal and transverse shear stresses or the maximum peak shear stress in any one direction, it is obviously more important to know the *resultant* maximum peak shear stress in order to accurately determine the stress levels sufficient to cause tissue damage, particularly ulceration of the insensitive diabetic foot. The *impulse* is another quantity which is thought to be particularly informative since it provides an indication of the magnitude of the shear stress combined with the duration of weight-bearing, which may be relevant to tissue damage. This is a quantity not calculated by previous workers studying plantar shear and unfortunately was not calculated here either because of the time-consuming nature of this analysis. However, in order to understand the process of plantar ulceration itself it is also necessary to consider how plantar shear is generated using measured traces in support. In this respect, it is apparent from a review of the literature that workers who have recorded in-shoe plantar shear stresses have not considered the patterns of shear generated in relation to shod foot function.

7.3.3 resultant maximum peak shear stresses

Resultant maximum peak shear stresses (rmpss) were calculated as vector-additions of the longitudinal and transverse components of shear (see equation 5.1, section 5.4.3) for measurements of plantar shear made beneath the medial four metatarsal heads and heel. The resultant directions of shear on the transducer are illustrated in figure 7.1.

There was found to be no significant difference in the magnitudes of *rmpss* between the two groups studied for the two conditions that were tested. However, with shear generated by a number of different mechanisms, for example movement of the underlying bony anatomy and deformation of the subcutaneous tissues, it is not possible to determine whether there were any similarities in respect of these many mechanisms between the two groups. In an analysis of the data gathered by **Pollard (1984)** in a study of similar groups, the magnitudes of shear stress were also found to

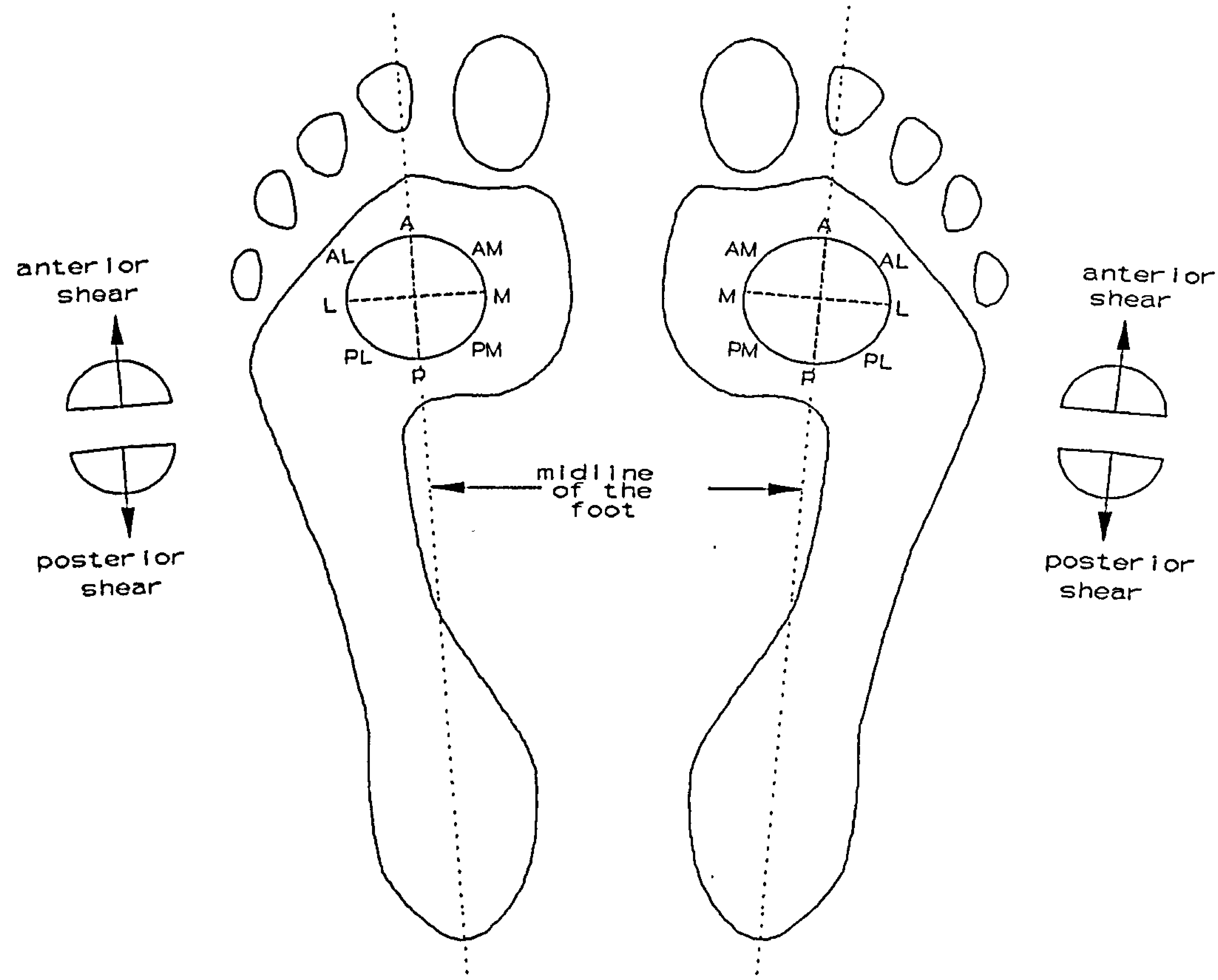

Figure 7.1. Resultant directions of shear on the transducer defined with respect to the midline of the foot: A - anterior; P - posterior; M - medial; L - lateral.

be similar between subjects and patients. *Peak-to-peak* longitudinal shear stresses[40] were measured beneath the first, second/third and fourth metatarsal heads (p=0.053, p=0.21 and p=0.46, respectively; subjects: n=10, patients: n=6, *op cit*). Although **Pollard's** is the only other comprehensive study of in-shoe plantar shear stresses to date, it is not possible to make a more quantitative comparison with the data gathered in the present study because of the differing methods of data analysis that were chosen: **Pollard** derived *peak-to-peak* values from measurements of shear stress. Although it was possible to analyse the data gathered in the present study using the method chosen by **Pollard** for comparison, this was not done because *peak-to-peak* values of shear stress are not thought to be particularly informative.

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40 **Pollard (1984) did not record transverse shear stresses in his patient group.**

7.3.4 waveforms of shear generated during each gait cycle

The traces in each shear record exhibit a series of repeating *waveforms*, each one generated during a single gait cycle (see Appendices F and G). Consistent waveforms are generally found in the shear records from one foot of an individual for each condition tested, ie. not wearing hosiery and wearing nylon hold-ups; and for the repeat measurements which were made three months apart. However, there are obvious differences in the patterns of shear generated between each site beneath the foot of an individual; between the same sites beneath contralateral feet; and between the same sites beneath the feet of different individuals. These differences could have arisen from differences in loading between adjacent plantar sites and between contralateral feet, and different gait patterns between individuals all giving rise to differences in the rate and direction of distortion of the plantar soft tissues and displacement of the underlying bony anatomy.

The pattern of a waveform may be described by its direction and form. In this respect a waveform may be either uni- or bi-directional, ie. either above *or* below, or above *and* below the datum line of the trace, respectively; and is either uni-, bi-, tri- or multi-phasic, ie. possessing either one, two, three or more distinct peaks, respectively (figure 7.2). In two previous studies of local plantar shear, a "sample set of traces" and a "typical set of waveforms" were presented **(Pollard, 1984, Tappin and Robertson, 1991)**. These waveforms were either uni-directional and uni-phasic or bi-directional and bi-phasic. In the shear records of the present study there are waveforms that fall into these two categories. However, there are also waveforms in the shear records which have a variety of other direction/form combinations. The method of flush mounting transducers into an inlay in the present study is thought to have facilitated the measurement of higher frequency events which were missed by other workers who taped transducers directly to the foot - the response of transducers taped to the foot is likely to be damped by the deformation of surrounding soft tissue during walking. As a result the records presented by other workers cannot be justifiably regarded as *typical*.

Based on the typical shear output from a forceplate (see figure 2.19), the direction of local plantar shear generated beneath the metatarsal heads was expected to be

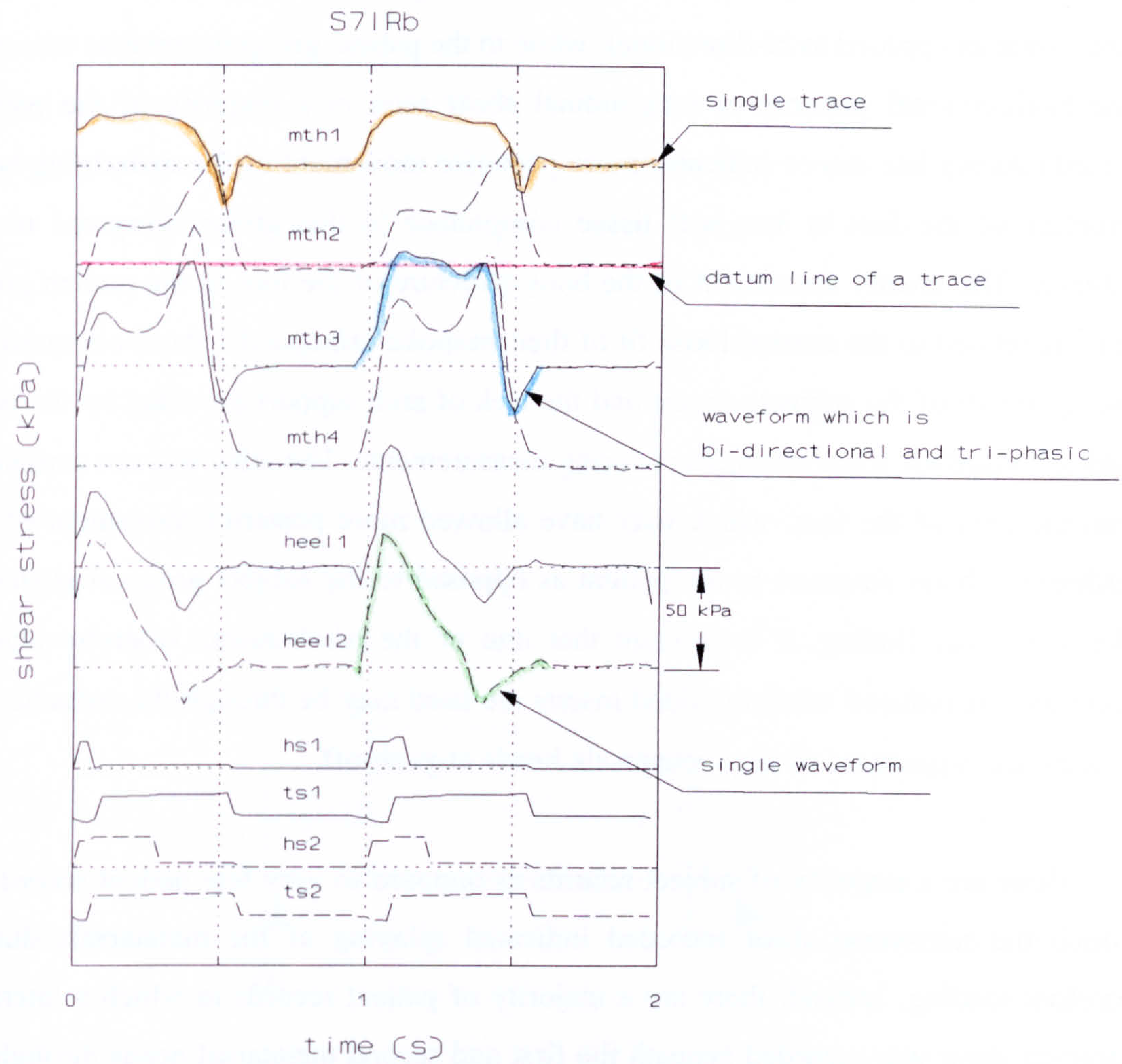


Figure 7.2. Terms used to describe the features in the shear records.

antero-medial between foot-flat and midstance and postero-lateral at push-off (relative to the transducer). The patterns of local plantar shear generated must on aggregate reflect deceleration of the body between foot-flat and midstance and acceleration at push-off. Locally, however, there were patterns of shear generated which were contrary to that expected globally. It is apparent from a simple visual inspection of each shear record that there were traces in some which indicated a local anterior shear at push-off. In these records, a posterior shear was simultaneously generated at other forefoot sites indicating twisting about the forefoot at push-off. Some forceplates are designed to measure torque, which would enable this twisting motion at push-off to be seen globally.

In a majority of subject shear records, the longitudinal shear recorded was uni-directional as opposed to bi-directional; while in the patient group the reverse was true. The bi-directional patterns of longitudinal shear seen in a majority of the patient records during late stance indicates more posterior movement of the underlying bony structure of the foot or less soft tissue compliance in this group compared to the subjects. The greater movement of the bony structure of the foot in the patient group may be related to the normal loose fit of their bespoke orthopaedic shoes compared to the tighter fit of the subjects shoes; and the lack of arch support provided by the inlay which housed the shear transducers during measurements. The inlay was not contoured into the arch of the foot, which may have allowed more posterior movement of the underlying bony structure in the patient as opposed to the subject group at push-off. Based on this finding, it is thought that one of the mechanisms whereby plantar ulceration is reduced when moulded inserts are used may be through the reduction of a posterior movement of the metatarsals heads at push-off.

There are a majority of subject records as opposed to very few patient records in which the transverse shear recorded indicated splaying of the metatarsals during forefoot loading. Instead, there are a majority of patient records in which a laterally directed shear was recorded beneath the first and second metatarsal heads throughout the stance phase, which peaked at push-off. This finding suggests that load was transferred more medially across the forefoot of the patients compared to the subjects prior to push-off (figure 7.3) - a view supported by the pressure data and the line of progression of the locus of the CoP in-shoe which was observed beneath the forefoot (see Chapter 5, tables 5.8 and 5.9). With a majority of healed ulcers noted beneath the medial side of the forefoot in the patient group, it is yet more apparent that a bias towards medial loading of the forefoot may be a significant risk factor in ulceration.

An inspection of the patient shear records revealed no specific pattern of shear was evident beneath the site of a healed ulcer. However, since shear may result in the generation of high stresses within the plantar soft tissue (**Zhang and Roberts, 1993**), it is clear that there may be a number of mechanisms by which this occurs. Constant sliding of the foot in a loose fitting or poorly fastened shoe during walking may be one mechanism leading to plantar soft tissue damage and eventual ulceration. Twisting

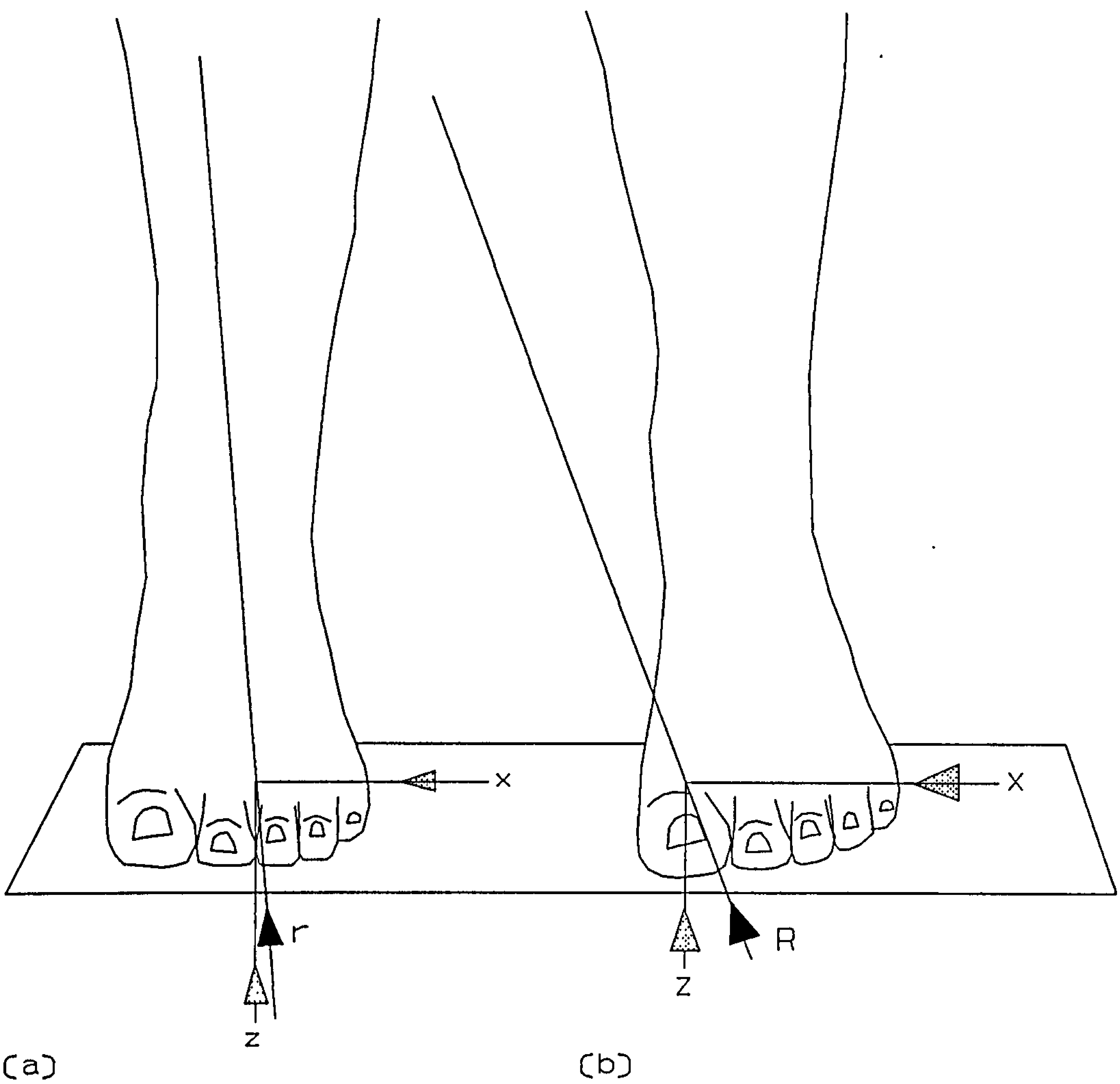


Figure 7.3. Generation of laterally directed shear beneath (a) the lesser and (b) all five metatarsal heads at push-off, the latter indicative of a more medial transfer of load.

in the region of the forefoot is, however, thought to be particularly destructive at a site where the soft tissue is placed in tension. In patients with healed ulcers, twisting is thought particularly likely to lead to re-ulceration as a result of the opposing movement of the scar tissue and surrounding soft tissue.

7.4 IN-SHOE PLANTAR SHEAR

7.4.1 generation of in-shoe plantar shear

The shear forces that are generated between the shoe outsole and ground can be transmitted to the foot as either shear or pressure on the plantar and dorsal aspects. Although in-shoe plantar shear arises from whole body ground reaction forces, its distribution may be determined by a combination of local factors such as: articulations of the underlying bony structure upon weight-bearing; deformation of the plantar soft tissues; and sliding of the foot relative to the shoe insole. When correctly fitting shoes are worn which hold the foot firmly into the rear of the shoe, the generation of plantar shear is thought to be largely dependent on the first factor, particularly during flattening and splaying of the metatarsals. In this case, the magnitudes of the local *rmpss* that are generated may be expected to be higher for a mobile as opposed to a rigid foot.

The patterns of plantar shear stress that are generated locally will be more variable than the whole body shear forces resulting from their integration. However, just as the whole body shear forces in the antero-posterior and medio-lateral directions reflect acceleration and deceleration of the body during various stages of the gait cycle, the patterns of local plantar shear must on aggregate reflect this also. It may be recalled from Chapter 2, section 2.5.7, that detailed descriptions were provided of the movements of the foot and bony anatomy during the stance phase and the consequent patterns of plantar shear that are expected to be generated beneath the metatarsal heads and heel. The patterns of shear that were described at these few sites gave a similar indication of acceleration and deceleration during the stance phase, which are indicated in the familiar patterns of the whole body shear forces. It is interesting to note then that there is only one shear record, among those of the subjects, in which the waveforms recorded closely resemble those expected from the overall forces recorded on a forceplate (*S4(lt)Rb*). This particular record is described in detail in the following section, which discusses the interaction between the foot and the shoe during walking.

7.4.2 interaction between the foot and the shoe during walking inferred from a subject shear record

An extract from the shear record of subject number 4 (*S4(lt)Rb*) appears in figure 7.4 and is discussed below with respect to the phases of gait. Recall first from Chapter 4, section 4.3.3, that two walks were necessary to record plantar shear stresses beneath the medial four metatarsal heads. Although the heel-switch outputs of these two walks are synchronised at heel-lift (line *D*), initial toe-to-ground contact occurred at different

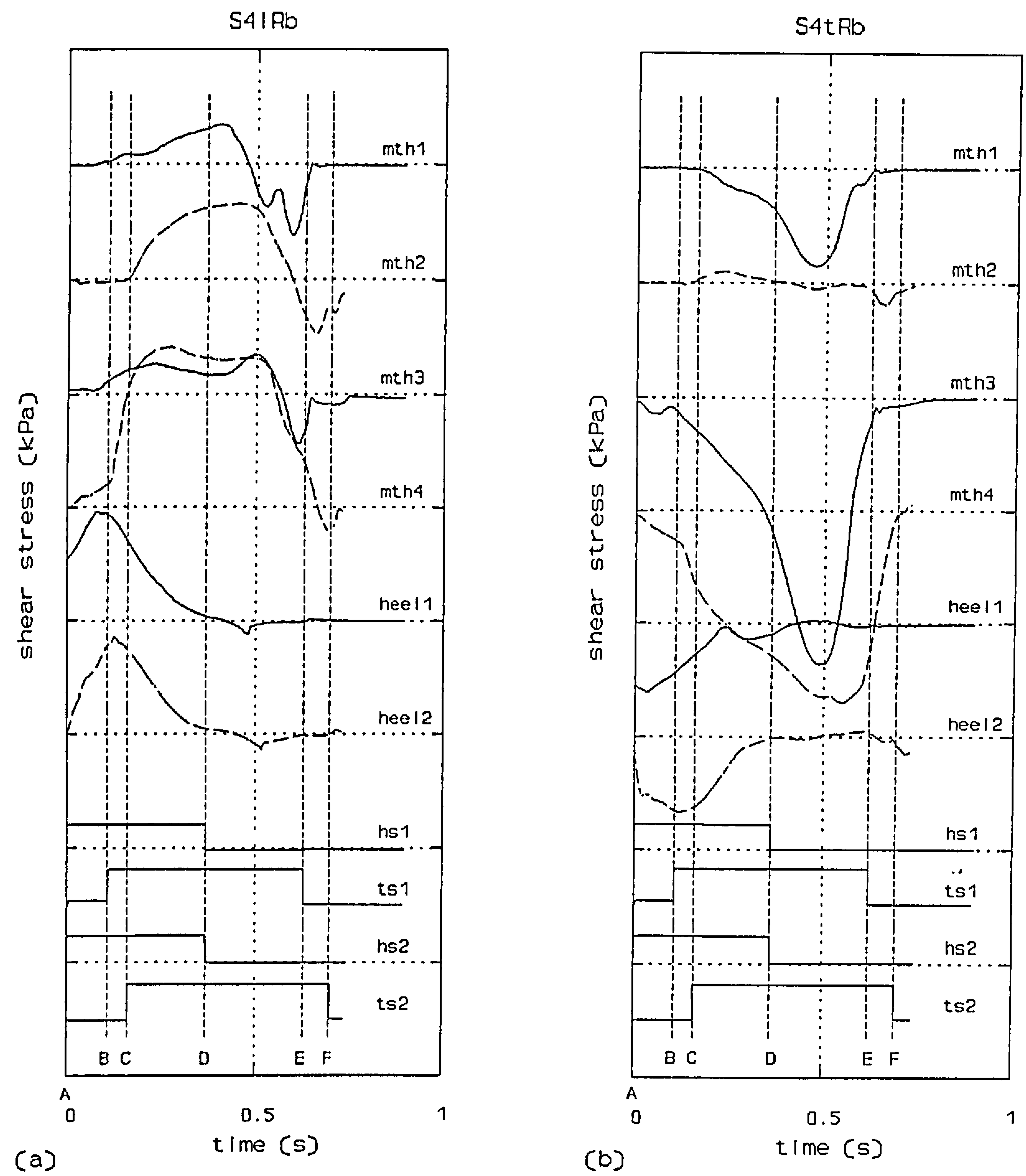


Figure 7.4. Extract from the right foot shear record of Subject number 4 - the waveforms from the second complete step are presented: (a) longitudinal and (b) transverse shear.

times: the lines *B* and *C* do not coincide. It is therefore important to remember that the waveforms recorded beneath the first and third metatarsal heads are not exactly synchronised with those recorded beneath the second and fourth metatarsal heads. As a result, any interpretation of this shear record as indicative of particular movements of the foot and underlying bony structure must be made bearing this in mind. Also, it is apparent from the footswitch data in figure 7.4 that the duration of the stance phase of these two walks was not similar: the lines *E* and *F* do not coincide.

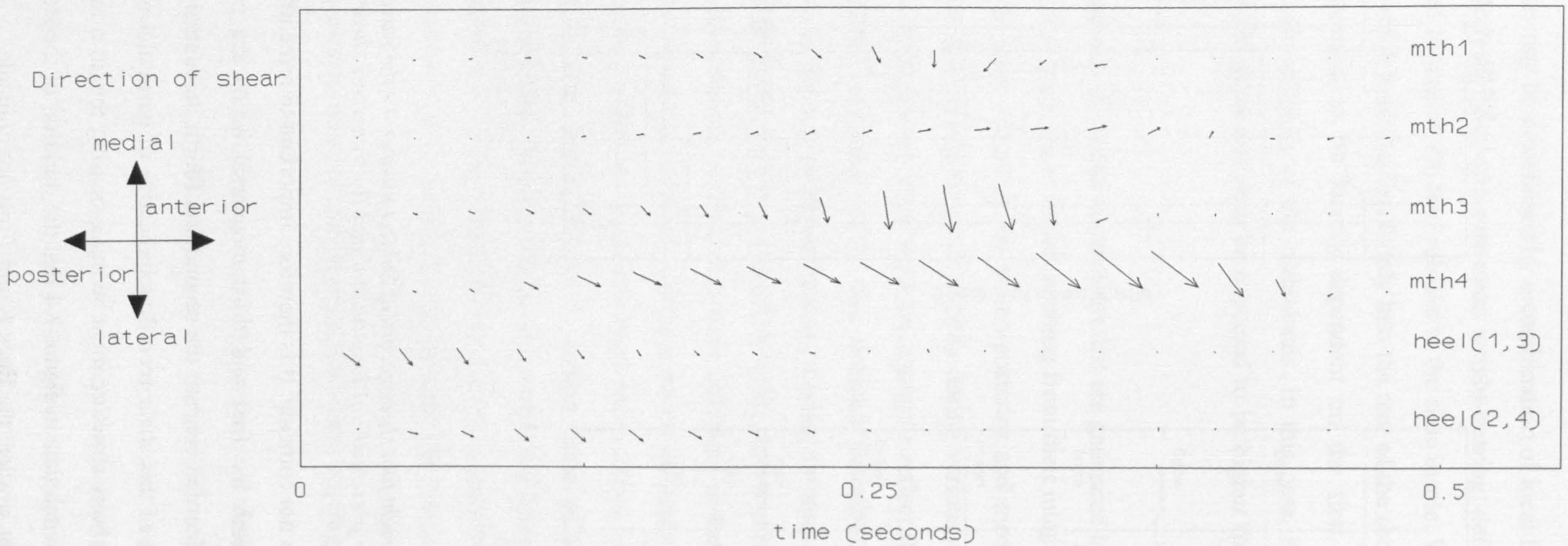


Figure 7.5. Vector diagram illustrating the magnitude and direction of the local rmpss for the waveforms of longitudinal and transverse shear presented in figure 7.4.

Early stance (walk 1: *A* to *B*, and walk 2: *A* to *C*). During this period an antero-lateral deceleration shear was recorded beneath the heel and lateral side of the forefoot (third and fourth metatarsal heads) (figure 7.5).

The direction and increasing magnitude of the shear recorded beneath the heel is coincident with the body being decelerated during this period. Because the forepart of the shoe outsole is not yet in contact with the ground, the shear recorded beneath the forefoot will be dependent on the shoe fit to maintain contact between the plantar skin and shoe insole. When a correctly fitting lace-up style shoe is worn, pressure will be exerted over the joint and instep which will reduce forward sliding of the foot and hence plantar shear compared to that for a loose fitting shoe (figure 7.6).

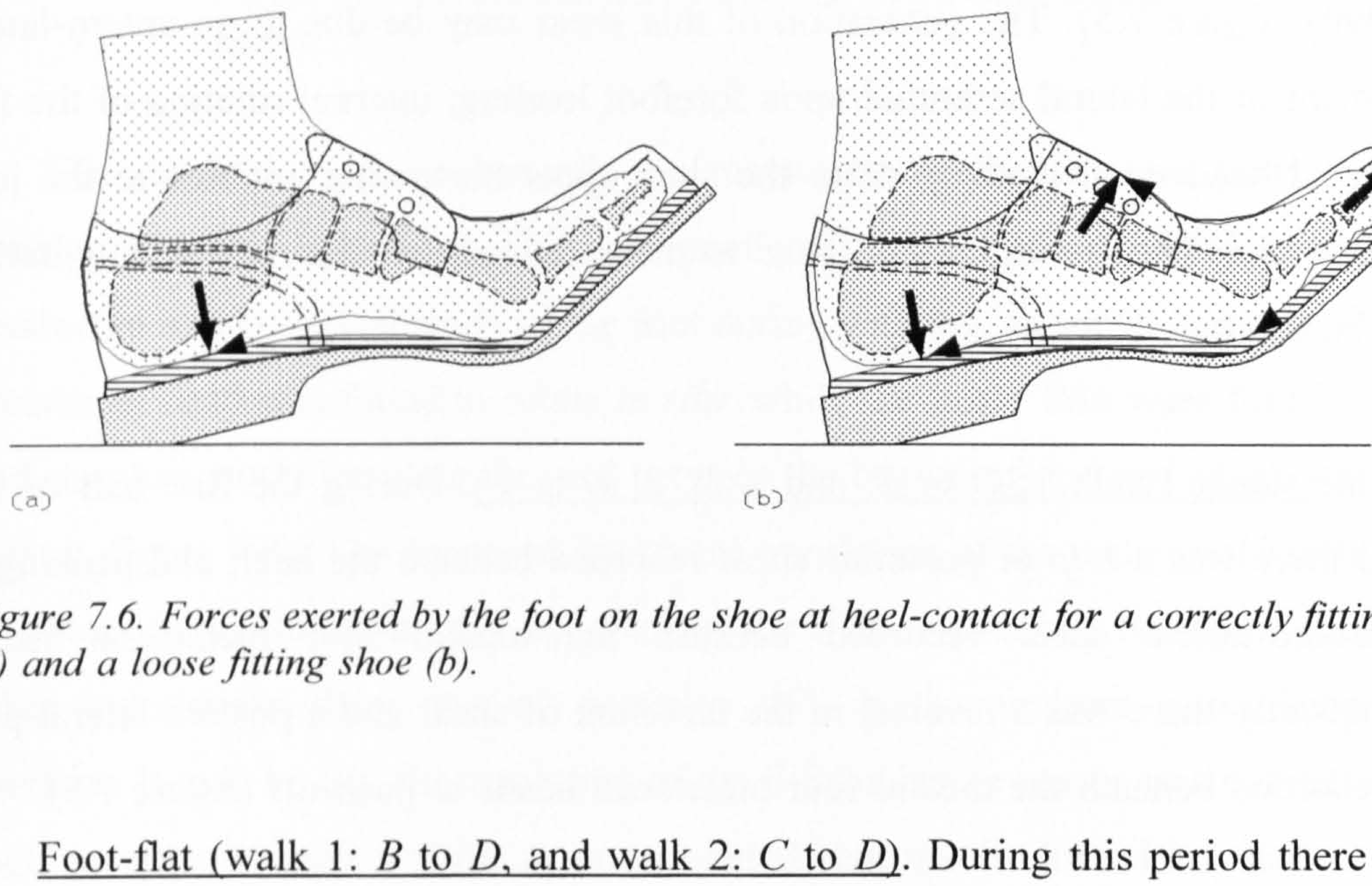


Figure 7.6. Forces exerted by the foot on the shoe at heel-contact for a correctly fitting shoe (a) and a loose fitting shoe (b).

Foot-flat (walk 1: *B* to *D*, and walk 2: *C* to *D*). During this period there was a gradual decrease in shear beneath the heel; an increase in the antero-lateral shear beneath the third and fourth metatarsal heads; and onset of shear beneath the medial two metatarsal heads - an anterior shear beneath the second metatarsal head and an antero-lateral shear beneath the first metatarsal head.

During the period of foot-flat the forefoot is gradually loaded as the body moves forward over the supporting foot. The plantar skin will tend to remain in the position at which it first comes into contact with the shoe insole when the forepart of the shoe outsole has made ground contact; and the underlying bony structure will distort, ie. the longitudinal and transverse arches will flatten and the metatarsals splay **(Lord et al, 1992)**. Again, depending on the shoe fit these movements and hence plantar shear may be more or less reduced.

In this particular subject the directions of shear recorded beneath the forefoot indicate flattening and lateral splaying of the third and fourth metatarsals: the magnitude of the longitudinal and transverse components of shear indicate a greater anterior and lateral movement of the fourth metatarsal head. It is particularly interesting to note the lack of transverse shear recorded beneath the second metatarsal head, which is considered to indicate the flattening of this metatarsal in the sagittal plane upon forefoot loading. The antero-lateral shear beneath the first metatarsal head is more surprising since splaying about the second metatarsal is expected **(Shereff et al, 1990)** (figure 7.5). The generation of this shear may be due to an antero-lateral movement of the lateral sesamoid upon forefoot loading; internal rotation of the first metatarsal head upon loading; or to the shoe upper narrowing anterior to the joint region forcing the first metatarsal head to move laterally as the medial longitudinal arch flattened.

Late stance (walk 1: *D* to *E*, and walk 2: *D* to *F*). During the first half of this period there was a *blip* of posterior shear recorded beneath the heel; and peaking of the antero-lateral shear recorded beneath the medial four metatarsal heads. Subsequently there was a reversal in the direction of shear and a postero-lateral peak was recorded beneath the medial four metatarsal heads at push-off (figure 7.5).

The *blip* of posterior shear recorded beneath the heel after heel-lift suggests a backward movement of the heel against the counter of the shoe. During late stance there is also a tendency for the heel to lift out of the shoe and a reversal of all plantar shear stresses (figure 7.7).

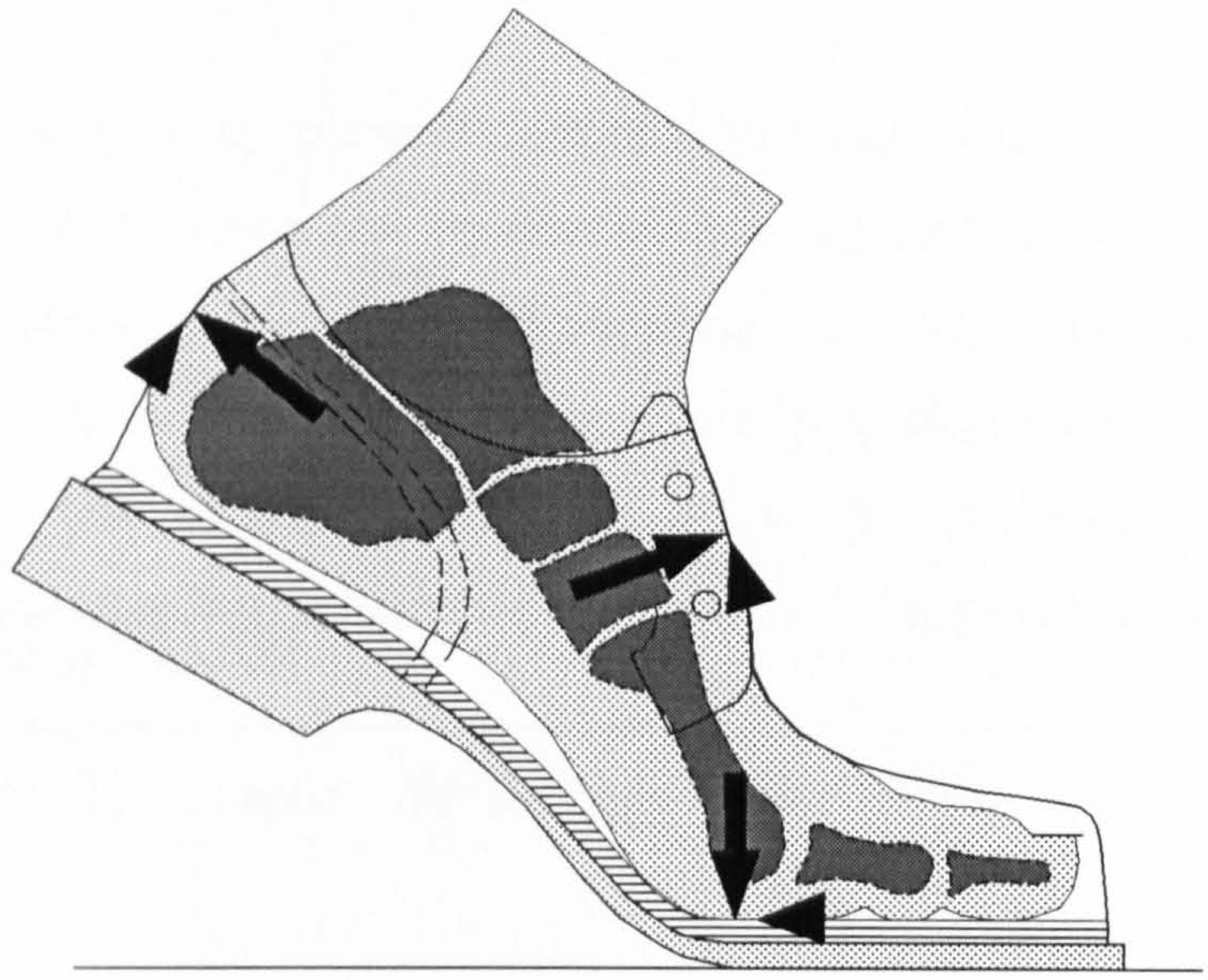


Figure 7.7. Forces exerted by the foot on the shoe during late stance.

The reversal in shear beneath the metatarsal heads indicates their backward translation during late stance. This is contradictory to the findings of **Price (1959)** who conducted a radiological study of the foot during walking, although barefoot. The first metatarsal head was found to rotate *in situ*; while the lesser four were found to rotate and simultaneously translate forward between the beginning and end of metatarsal rise (*op cit*, figure 7.8). The proposed backward translation of the metatarsal heads during late stance was, however, not apparent in the F-Scan pressure record of this or the other individuals: there was no posterior shift in the pressure peaks beneath the forefoot. It may be that the resolution of the F-Scan insole sensor was not sufficiently high to register such a shift. It is concluded that the backward translation of the metatarsal heads during late stance was very small, but a large posterior shear was generated in some individuals due to a combination of high pressure beneath the ball and a small translation of the metatarsal heads being transmitted through tensed soft tissue - the tension in the soft tissue being brought about by extension of the metatarsophalangeal joints at this time (*op cit*, figure 7.9).

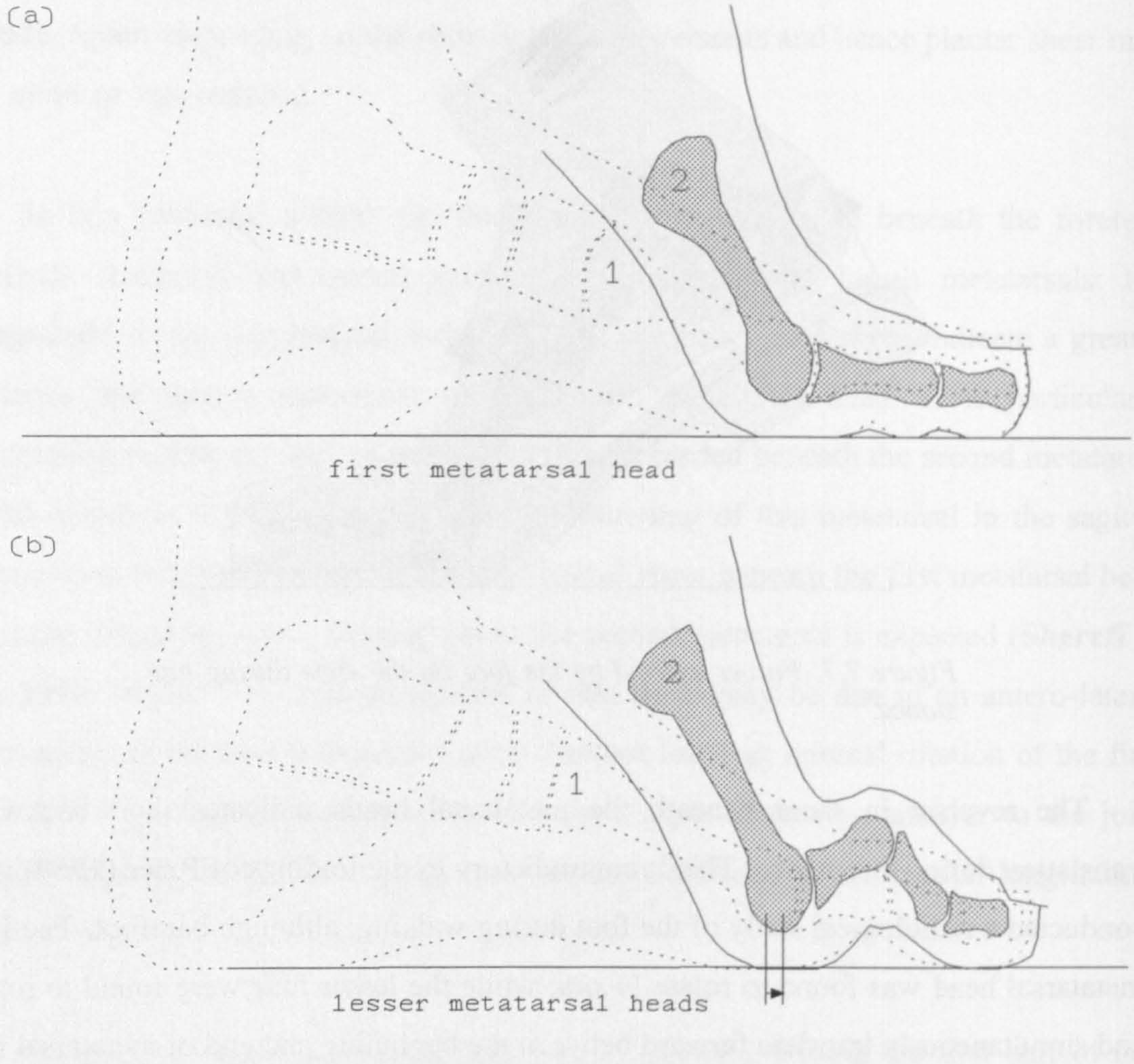


Figure 7.8. Motion of the first (a) and lesser (b) metatarsal heads during gait, as observed by Price (1959).

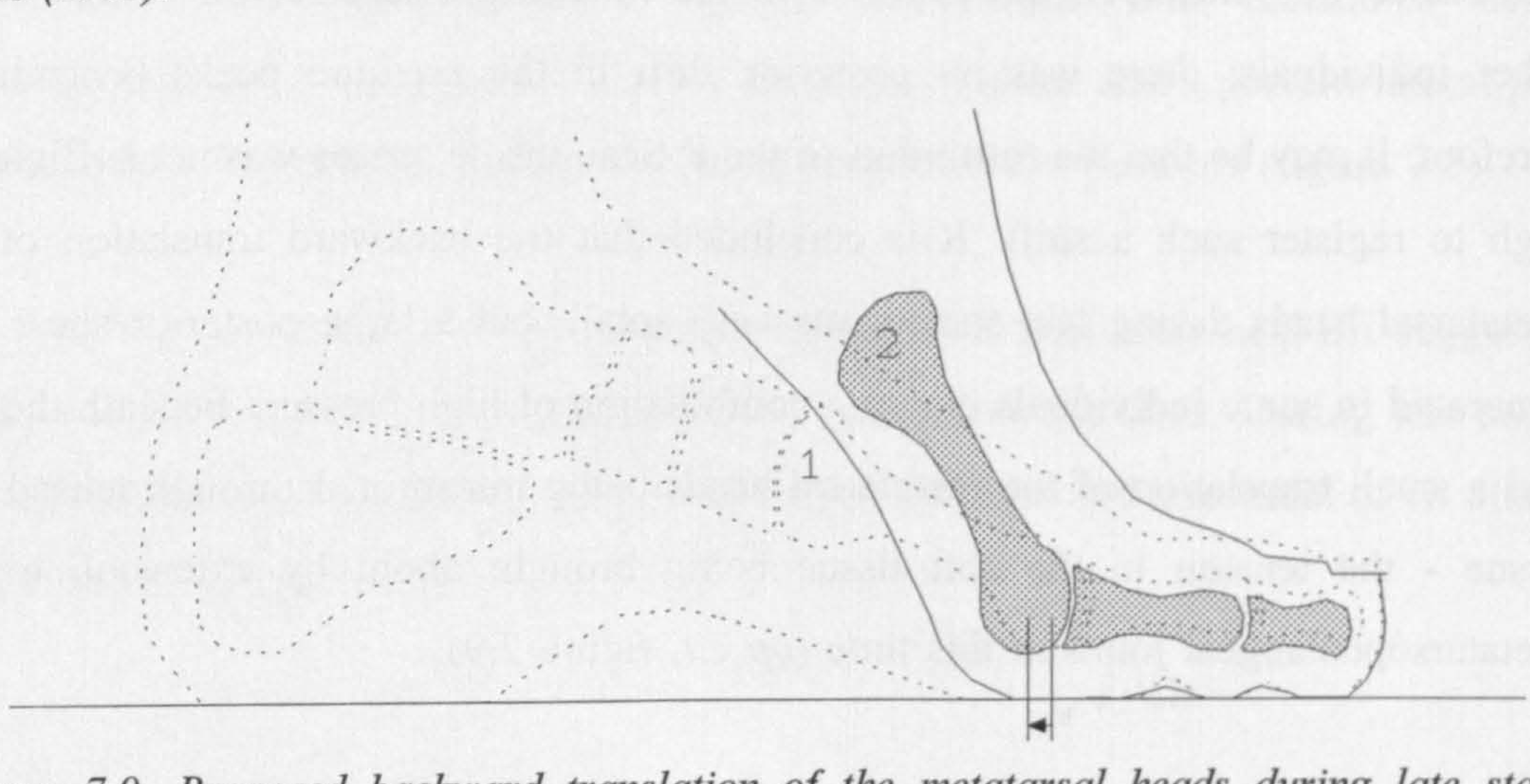


Figure 7.9. Proposed backward translation of the metatarsal heads during late stance generating a posteriorly directed shear at push-off (illustrated for the first metatarsal).

7.5 SUMMARY

There are obviously many personal factors which may affect the magnitudes and patterns of local plantar shear that are generated during walking, for example walking speed. This Chapter has also highlighted the possible effects of shoe fit on the patterns and magnitudes of local plantar shear that are generated. It is apparent that an understanding of how local plantar shear is generated is particularly relevant in the design of shoewear for patients with diabetes who are predisposed to plantar ulceration.

To conclude this Chapter, the findings of the present study will be briefly summarised.

In agreement with the only other study of in-shoe plantar shear, similar magnitudes of *rmpss* were measured in the two groups of the present study.

The shear data recorded indicated load was transferred more medially across the forefoot of the patients, as opposed to the subjects, at push-off. This finding is consistent with a number of other observations made in the patient group: a majority of patients were assessed to have a forefoot valgus condition; the pressure data recorded indicated a medial bias in forefoot loading; and there was a high incidence of healed plantar ulcers beneath the first and second metatarsal heads.

The patterns of longitudinal shear seen in late stance indicated more posterior movement of the metatarsals or less soft tissue compliance in the patient compared to the subject group.

In the patient group, there were a majority of cases in which the site of a healed ulcer was coincident with the site of highest *rmpss* recorded beneath the forefoot.

In an analysis of the shear and biomechanical data recorded, there was no obvious association between the magnitudes and patterns of local plantar shear generated and the ranges of joint motion that were measured, for the two groups studied.

CHAPTER 8
GENERAL DISCUSSION

8.1 INTRODUCTION

Although devices for measuring plantar stresses have been available for 70 years (**Frostell, 1925**, cited in **Horan, 1976**), their use in investigating neuropathic plantar ulceration is only recent: **Bauman et al (1963)** were the first to study pressures and **Pollard et al (1983)** shear stresses in patients with healed plantar ulcers. In many studies of in-shoe plantar stresses utilising discrete transducers, measurements were made with the transducers taped to the sole of the foot. However, the interposition of a rigid object between the foot and supporting surface will undoubtedly have a perturbing effect on the magnitudes of local plantar stresses that are generated. Although **Pollard (1984)** acknowledged that measurements made with transducers taped to the sole of the foot may overestimate the magnitudes of shear stress generated, all workers measuring shear have since used this methodology. In contrast, workers studying plantar pressures have tried to overcome the inaccuracies introduced by this methodology by developing inlays in which to flush-mount their transducers. Additionally, inaccuracies may have been introduced into measurements of both pressure and shear as a result of many workers relying on static calibration of their transducers prior to making dynamic measurements.

During this study of in-shoe plantar stresses and their role in neuropathic ulceration, measurements of shear were made using a transducer developed by **Williams (1993)** in the Department of Medical Engineering & Physics at King's College Hospital (Dulwich). An inlay was developed into which the transducers could be flush mounted to minimise measurement errors due to perturbation; and the transducers were quasi-statically calibrated both before and after clinical measurements. In-shoe plantar shear stresses were recorded beneath the feet of asymptomatic subjects and patients with diabetes who had a history of neuropathic ulceration: measurements were made in regions of peak pressure which were found by pressure mapping.

8.2 EVALUATION OF THE SHEAR TRANSDUCER SYSTEM

During this study, measurements of shear were made using a bi-axial transducer. In contrast, **Pollard (1984)** made measurements using a uni-axial transducer, which required the patient to make repeated walks in order to record longitudinal and transverse shear stresses. Repeated walks were also necessary in the present study, but for a different reason. Because of the diameter of the housing which was required to accommodate and locate the shear transducer in the inlay, it was not possible to place transducers in close proximity beneath the forefoot. Consequently, local plantar shear stresses could not be recorded beneath adjacent metatarsal heads during a single walk. It is obviously preferable to mount more transducers into an inlay at once in order to gather data from more sites at a time, but this would require the development of a smaller transducer. Such instrumentation development based on the magneto-resistive principle is not thought to be potentially rewarding because of possible difficulties in device fabrication and assembly as a result of working to close tolerances. Instrumentation development based on piezoelectric technology is seen as a viable alternative, although there is an associated high cost due to the necessary interfacing electronics. Workers at the University of Kent have recently developed a biaxial shear transducer from PVdF and copolymer films (**Akhlaghi and Pepper, 1993**), but to date there have been no publications of clinical measurements.

8.3 IN-SHOE PLANTAR STRESS

The waveforms in the shear records presented in the Appendices are of the actual output from each shear transducer axis - for clarity, the longitudinal and transverse components of shear were presented separately. An alternative and perhaps more informative form of data presentation is that of the vector diagram in which the local *resultant* shear stresses are plotted, similar to the butterfly diagrams of normal force (**Tait and Rose, 1979**). By presenting shear data in this way, the processes whereby local plantar shear is generated may be more readily apparent. Vector diagrams of local plantar shear were, however, not presented here because of a combination of time constraints and the considered necessity to present the actual waveforms of shear recorded.

Local *resultant maximum peak shear stresses (rmpss)* were evaluated by vector-addition of the two mutually perpendicular components of shear recorded. With pressure also recorded during this study it is seemingly possible to evaluate the *maximum peak stress* generated beneath each of the medial four metatarsal heads and heel during gait. It is apparent that a knowledge of the local maximum peak stresses generated is more important than maximum peak values of individual components of stress, particularly when seeking to determine the actual magnitudes of stress that are sufficient to cause tissue damage. Unfortunately, pressure and shear data could not be recorded simultaneously during this study. Consequently, it was not possible to instantly determine the resultant combined stress generated at each site. Alternatively, the pressure and shear data could be standardised in order to calculate local maximum peak stresses. However, seeing as the pressure and shear data were recorded on separate days, this method of analysis may result in inaccuracies due to possible differences in gait pattern between measurement days.

At the outset of this study the magnitudes of *rmpss* and waveforms of shear were expected to differ significantly between the two groups. This hypothesis was based on the results of previous studies which found significantly higher plantar pressures **(Stokes et al, 1975, Ctercteko et al, 1981, Boulton et al, 1984, Veves et al, 1991)** and significantly reduced ranges of joint motion in the feet of patients with diabetes compared to healthy subjects **(Delbridge et al, 1987, Cavanagh et al, 1991a, Fernando et al, 1991)**. In addition, significant differences in the magnitudes of *rmpss* and waveforms of shear were expected due to the looser fit of the patients shoes as a result of their normal inserts being replaced by a thinner inlay. However, on statistical analysis, the magnitudes of *rmpss* measured were found to be similar at corresponding sites between the groups, although differences were apparent in the waveforms of shear recorded. On visual inspection of the shear records, there was a greater proportion from the patient group which had bi-directional waveforms, suggesting more movement of the underlying bony structure of the foot during late stance or less soft tissue compliance compared to the subjects.

In a comparison between the waveforms of shear recorded in this study with those recorded by **Pollard (1984)**, it is thought possible to conclude that the method of taping shear transducers to the sole of the foot may not favour the measurement of high frequency events, mainly due to soft tissue deformation hindering the sliding mechanism of the transducer. The waveforms in the "sample set of traces" presented by **Pollard** indicate less movement of the foot and underlying bony structure compared to those recorded here for similarly matched subjects. The waveforms presented by **Pollard** were either uni-directional and uni-phasic or bi-directional and bi-phasic. In the shear records of the present study, although there were waveforms that fell into these two categories, there were also waveforms which had a variety of other direction/form combinations.

Variability in the waveforms of plantar shear which were noted between individuals are thought to be largely dependent on shoe fit. Shoe fitting is a complex issue and the fitting factors which affect plantar shear are difficult to define. However, it can be generalised that shoes which are either correct or tight fitting will reduce bony articulations, and hence plantar shear, compared to a loose fitting shoe. New leather shoes tend to soften with wear and so even a perfectly fitted shoe will never eliminate plantar shear. In fact with the softening of leather shoes in use, it is possible that plantar shear may be greater in an old compared to a new pair of shoes. It is apparent that whereas shoe fit may have a significant effect on the patterns and magnitudes of in-shoe plantar shear and even the patterns of plantar pressure that are generated, it may not have any effect on the magnitudes of peak pressure. Instead, the magnitudes of peak pressure will be affected by factors such as insole cushioning and plantar soft tissue thickness.

8.4 PLANTAR ULCERATION OF THE INSENSITIVE DIABETIC FOOT

The role of pressure and shear in the formation of plantar ulcers is complicated by the fact that there is probably more than one way in which these stresses act together to cause ulceration. Ulcers frequently develop beneath the forefoot of patients with diabetes at sites of heavy callus (**Edmonds, 1986**) and may be supposed to occur as a result of compliant tissue trapped between the metatarsal heads and callus being subjected to high local stresses (direct and shear) during weight-bearing (**Thompson, 1988**). The application of shear and pressure to soft tissue was investigated by **Zhang**

and Roberts (1993) in a theoretical analysis who concluded that there would be an increase in the internal stress distribution ahead of the point of shear application. Extrapolating this to the foot, during the period of foot-flat, anterior movement of the metatarsal heads relative to fixed plantar skin may occur and cause damage to the subcutaneous tissue anterior to a metatarsal head. Subcutaneous tissue damage leading to plantar ulceration may also occur from the application of abnormally high stresses generated by particular actions during walking, for example, twisting about the forefoot at push-off. In patients with healed forefoot ulcers, twisting may be particularly destructive, resulting in re-ulceration from high stresses generated at the interface between the hard scar tissue which penetrates the subcutaneous tissue.

The shoewear currently supplied to patients with diabetes who are at risk of ulcerating or re-ulcerating is designed to accommodate moulded or cushioning inserts which redistribute plantar pressure and reduce pressure peaks. Shear too presents a risk of plantar ulceration and may also be minimised through the use of moulded inserts: moulding into the medial longitudinal arch will result in the exertion of normal forces in this area and can be anticipated to reduce sliding of the foot in the shoe, as well as limiting articulations of the underlying bony structure by supporting the longitudinal arch. It is often the case that patients with diabetes often first ulcerate while wearing fashion shoes. If it is assumed, quite reasonably, that the fashion shoes worn by these patients are correctly fitted and are correctly worn, plantar shear may already be minimised. Hence initial plantar ulceration may occur through the major action of pressure generating a high internal stress distribution in the subcutaneous tissues. In fact fashion shoes tend to lack insole cushioning and with the addition of high heels direct pressure is increased beneath the forefoot (**Snow et al, 1992**).

8.5 CONCLUSIONS

During this research a methodology was developed for in-shoe plantar shear stress measurement; and in-shoe plantar shear stresses were investigated in asymptomatic subjects and patients with diabetes who had a history of plantar neuropathic ulceration.

Measurements of shear were made using a bi-axial transducer, which was flush mounted in a removable shoe inlay to minimise measurement errors due to perturbation. An evaluation of this methodology revealed individuals did not suffer any discomfort from or were not aware of the presence of the transducers during measurements. However, with this methodology the large diameter and design of the transducer, particularly with regard to the attachment of the signal leads, were found to be limiting factors in the number that could be placed in close proximity beneath the forefoot at any one time to instrument the medial four metatarsal heads.

In-shoe plantar shear stresses were measured beneath the medial four metatarsal heads and heel in regions of peak pressure located by pressure mapping. Both the *rmpss* and maximum peak pressures measured at corresponding sites were similar between the two groups studied. The similar magnitudes of *rmpss* measured between the two groups could not be related to similarities between the subjects and patients with regard to the mechanisms of shear generation. In contrast, the similar magnitudes of maximum peak pressure that were measured were thought to be due to the combined effects of considerable spatial smoothing of localised peaks, as a result of the support provided to the foot during measurements and the large area over which pressures were averaged. It is apparent from work by **Lord (1993)** that pressures should be averaged over a very small area in order to measure *true* maximum peaks and to determine any real differences that may exist between these two groups.

An analysis of the waveforms of shear recorded revealed considerable differences between individuals. With the chosen method of data presentation, ie. of the actual waveforms of shear recorded, it was noted that considerable time was necessary to visually analyse a shear record in order to relate this to the possible movements of the foot and underlying bony anatomy. As a result of this, the vector diagram was examined as an alternative form of data presentation. Vector diagrams have the advantage that both longitudinal and transverse components of shear may be presented together in a concise and meaningful way. However, vector diagrams necessarily have a poor resolution for clarity in contrast to the continuous form of data presented as waveforms. As a result, the poor resolution of the vector diagram may result in a loss of important functional data.

It is concluded that research into plantar shear stress measurement is still in its infancy, with regard to equipment and data analysis, to be of use in a clinical setting assessing patients with diabetes who are at risk of plantar ulceration. It is apparent that with numerous devices currently available for plantar pressure measurement and the present understanding of the data that may be gathered, these measurements are much more immediately useful clinically.

8.6 SUGGESTIONS FOR FURTHER RESEARCH

There are two possible approaches to extending this work, aimed at increasing the present understanding of plantar ulceration. Firstly, it is clearly desirable to conduct a prospective study of in-shoe plantar shear in patients with diabetes who have not ulcerated, similar in design to the prospective study of **Veves et al (1992)** who investigated plantar pressure. Through an extended study such as this it would be possible to investigate the movements of the foot and bony anatomy, through shear records, which may lead to subcutaneous tissue damage and eventual plantar ulceration. Secondly, it is thought important to investigate the stresses in the subcutaneous tissues that result from pressure and shear applied to the plantar surface and which may lead to ulceration. In this respect, finite element (FE) analysis is thought to be the best approach. Further measurements of in-shoe plantar stresses are not thought to be necessary at this time for such a model. Instead an FE model would use data already available. At present, research is underway at the Penn State University, USA, to create a finite element model of the foot/shoe interface based on plantar pressure data and measures of the bony distortions of the foot as derived from weight-bearing radiographs **(Cavanagh, 1994)**. However, with the addition of shear data, such as that gathered in the present study, such models can be extended to provide more accurate information and be more informative.

APPENDIX A

Table A.I. Kolmogorov-Smirnov statistics for the data recorded.

Variable	*K-S* value	*P*-value
Age	0.713	0.689
Bodymass	0.378	0.999
Cadences recorded during pressure measurements	0.779	0.578
Duration of the stance phase recorded during pressure measurements	0.691	0.726
Duration of the swing phase recorded during pressure measurements	0.635	0.814
Duration of the double-support phase recorded during pressure measurements	0.845	0.473
maximum peak pressures recorded beneath the hallux	0.751	0.625
maximum peak pressures recorded beneath the first metatarsal head	0.719	0.680
maximum peak pressures recorded beneath the second metatarsal head	0.561	0.911
maximum peak pressures recorded beneath the third metatarsal head	0.718	0.682
maximum peak pressures recorded beneath the fourth metatarsal head	1.068	0.204
maximum peak pressures recorded beneath the heel	0.907	0.383
Timings of onset of pressure beneath the hallux	0.480	0.975
Timings of onset of pressure beneath the first metatarsal head	0.956	0.320
Timings of onset of pressure beneath the second metatarsal head	0.760	0.610
Timings of onset of pressure beneath the third metatarsal head	0.888	0.409
Timings of onset of pressure beneath the fourth metatarsal head	0.641	0.805
Timings of onset of pressure beneath the heel	1.290	0.072

Table A.I (continued). Kolmogorov-Smirnov statistics for the data recorded.

Variable	*K-S* value	*P*-value
Timings of peak pressure beneath the hallux	0.792	0.558
Timings of peak pressure beneath the first metatarsal head	0.759	0.613
Timings of peak pressure beneath the second metatarsal head	0.541	0.931
Timings of peak pressure beneath the third metatarsal head	0.691	0.726
Timings of peak pressure beneath the fourth metatarsal head	0.846	0.472
Timings of peak pressure beneath the heel	0.700	0.712
Timings of cessation of pressure beneath the hallux	1.003	0.266
Timings of cessation of pressure beneath the first metatarsal head	0.933	0.349
Timings of cessation of pressure beneath the second metatarsal head	0.775	0.586
Timings of cessation of pressure beneath the third metatarsal head	0.971	0.303
Timings of cessation of pressure beneath the fourth metatarsal head	0.580	0.889
Timings of cessation of pressure beneath the heel	0.523	0.948
Cadences recorded during shear measurements	0.591	0.876
Resultant maximum peak shear stresses recorded beneath the first metatarsal head (not wearing hosiery)	0.724	0.670
Resultant maximum peak shear stresses recorded beneath the second metatarsal head (not wearing hosiery)	0.793	0.556
Resultant maximum peak shear stresses recorded beneath the third metatarsal head (not wearing hosiery)	0.654	0.786
Resultant maximum peak shear stresses recorded beneath the fourth metatarsal head (not wearing hosiery)	0.602	0.861

Table A.I (continued). Kolmogorov-Smirnov statistics for the data recorded.

Variable	*K-S* value	*P*-value
Resultant maximum peak shear stresses recorded beneath the heel (simultaneous with measurements made beneath the first and third metatarsal heads) (not wearing hosiery)	0.449	0.988
Resultant maximum peak shear stresses recorded beneath the heel (simultaneous with measurements made beneath the second and fourth metatarsal heads) (not wearing hosiery)	0.781	0.575
Resultant maximum peak shear stresses recorded beneath the first metatarsal head (wearing nylon hold-ups)	0.867	0.439
Resultant maximum peak shear stresses recorded beneath the second metatarsal head (wearing nylon hold-ups)	0.653	0.788
Resultant maximum peak shear stresses recorded beneath the third metatarsal head (wearing nylon hold-ups)	0.842	0.478
Resultant maximum peak shear stresses recorded beneath the fourth metatarsal head (wearing nylon hold-ups)	0.632	0.820
Resultant maximum peak shear stresses recorded beneath the heel (simultaneous with measurements made beneath the first and third metatarsal heads) (wearing nylon hold-ups)	0.666	0.767
Resultant maximum peak shear stresses recorded beneath the heel (simultaneous with measurements made beneath the second and fourth metatarsal heads) (wearing nylon hold-ups)	0.635	0.815
Resultant maximum peak shear stresses recorded during repeat shear tests (subjects not wearing hosiery)	0.666	0.760
Resultant maximum peak shear stresses recorded during repeat shear tests (subjects wearing nylon hold-ups)	0.594	0.872
Resultant maximum peak shear stresses recorded during repeat shear tests (patients not wearing hosiery)	0.623	0.832
Resultant maximum peak shear stresses recorded during repeat shear tests (patients wearing nylon hold-ups)	0.460	0.984
Ankle dorsiflexion	0.729	0.578
Calcaneal inversion	0.808	0.532
Calcaneal eversion	0.462	0.983

APPENDIX B

Table B-I. Subject cadences (steps/minute) during plantar pressure measurements, calculated from the duration of each complete step in the pressure records analysed.

Subject	Left foot			Right foot		
	step 1	step 2	step 3	step 1	step 2	step 3
1	116.5	111.1	113.2	113.2	116.5	108.1
2†	111.1	115.4	111.1	113.2	112.1	115.4
3	113.2	112.1	111.1	110.1	114.3	-
4‡	104.3	109.1	106.2	106.2	107.1	-
5	112.1	113.2	112.1	112.1	111.1	113.2
6	120.0	115.4	116.5	118.8	115.4	-
7‡	117.6	114.3	114.3	113.2	115.4	115.4
8†	105.3	105.3	-	111.1	107.1	110.1
9†	96.0	95.2	-	100.8	99.2	-

† pressure recordings were not obtained from both feet simultaneously.
‡ analysis performed on the first set of in-shoe pressure recordings which were made.

Table B-II. Patient cadences (steps/minute) during plantar pressure measurements, calculated from the duration of each complete step in the pressure records analysed.

Patient	Left foot			Right foot		
	step 1	step 2	step 3	step 1	step 2	step 3
1	105.3	114.3	-	109.1	106.2	115.4
2	94.5	93.0	-	93.8	94.5	95.2
3‡+	111.1	109.1	99.2	-	-	-
4	105.3	100.0	-	106.2	104.3	96.0
5	115.4	117.6	116.5	118.8	116.5	120.0
6‡	80.0	76.4	-	76.9	-	-
7†	96.0	96.0	-	100.0	84.5	-

† pressure recordings were not obtained from both feet simultaneously.
‡ analysis performed on the first set of in-shoe pressure recordings which were made.
+ pressure recordings were not obtained from beneath the right foot.

APPENDIX C

Page

Table C-I. Durations of the stance, swing and double support phases (expressed as a percentage of each complete gait cycle) in the subject pressure recordings.

Subject	Stance (right foot)	Swing (right foot)	Stance (left foot)	Swing (left foot)	Double support (rhs-lto)	Double support (lhs-rto)
1	65.1	34.9	66.0	34.0	14.6	15.1
	64.1	35.9	62.0	38.0	13.0	16.3
	61.3	38.7	64.2	35.8	11.3	11.7
2†	59.4	40.6	61.1	38.9	-	-
	58.9	41.1	62.5	37.5	-	-
	60.6	39.4	61.1	38.9	-	-
3	58.7	41.3	60.4	39.6	10.4	10.1
	60.0	40.0	63.6	36.4	11.2	11.4
	-	-	60.2	39.8	10.2	-
4‡	62.8	37.2	63.5	36.5	13.0	12.4
	65.2	34.8	63.6	36.4	12.7	13.4
	-	-	64.6	35.4	13.3	-
5	62.6	37.4	62.4	37.6	11.0	13.1
	62.0	38.0	64.2	35.8	13.2	13.9
	63.2	36.8	63.6	36.4	11.2	15.1

† pressure recordings were not obtained from both feet simultaneously, as a result the duration of the double-support phase could not be calculated.
‡ analysis performed on the first set of in-shoe pressure recordings which were made.
rhs - right heel-strike; lto - left toe-off; lhs - left heel-strike; rto - right toe-off.

Table C-I (continued). Durations of the stance, swing and double support phases (expressed as a percentage of each complete gait cycle) in the subject pressure recordings.

Subject	Stance (right foot)	Swing (right foot)	Stance (left foot)	Swing (left foot)	Double support (rhs-lto)	Double support (lhs-rto)
6	61.4	38.6	62.0	38.0	11.0	12.9
	63.5	36.5	62.5	37.5	12.5	13.5
	-	-	63.1	36.9	12.6	-
7‡	60.4	39.6	61.8	38.2	9.8	10.4
	59.6	40.4	62.9	37.1	10.5	12.5
	60.6	39.4	62.9	37.1	11.4	12.5
8†	63.0	37.0	64.9	35.1	-	-
	60.7	39.3	64.0	36.0	-	-
	61.5	38.5	-	-	-	-
9†	63.0	37.0	64.0	36.0	-	-
	62.0	38.0	64.3	35.7	-	-

† pressure recordings were not obtained from both feet simultaneously, as a result the duration of the double-support phase could not be calculated.
‡ analysis performed on the first set of in-shoe pressure recordings which were made.
rhs - right heel-strike; lto - left toe-off; lhs - left heel-strike; rto - right toe-off.

Table C-II. Durations of the stance, swing and double support phases (expressed as a percentage of each complete gait cycle) in the patient pressure recordings.

Patient	Stance (right foot)	Swing (right foot)	Stance (left foot)	Swing (left foot)	Double support (rhs-lto)	Double support (lhs-rto)
1	61.8	38.2	60.5	39.5	11.4	12.7
	63.7	36.3	63.8	36.2	11.4	12.4
	62.5	37.5	-	-	-	14.4
2	67.2	32.8	68.5	31.5	18.1	17.2
	70.1	29.9	65.9	34.1	16.3	20.5
	74.6	25.4	-	-	-	23.0
3‡	-	-	61.1	38.9	-	-
	-	-	65.5	34.5	-	-
	-	-	62.8	37.2	-	-
4	63.7	36.3	67.5	32.5	16.7	15.0
	70.4	29.6	66.7	33.3	17.5	21.7
	64.0	36.0	-	-	-	15.2
5	64.4	35.6	63.5	36.5	12.5	13.9
	66.0	34.0	64.7	35.3	15.7	15.5
	65.0	35.0	63.1	36.9	13.6	13.0

‡ analysis performed on the first set of in-shoe pressure recordings which were made.
rhs - right heel-strike; lto - left toe-off; lhs - left heel-strike; rto - right toe-off.

Table C-II (continued). Durations of the stance, swing and double support phases (expressed as a percentage of each complete gait cycle) in the patient pressure recordings.

Patient	Stance (right foot)	Swing (right foot)	Stance (left foot)	Swing (left foot)	Double support (rhs-lto)	Double support (lhs-rto)
6‡	62.8	37.2	65.3	34.7	14.7	15.4
	-	-	64.3	35.7	12.1	-
7†	65.0	35.0	66.4	33.6	-	-
	64.1	35.9	66.4	33.6	-	-

† pressure recordings were not obtained from both feet simultaneously, as a result the duration of the double-support phase could not be calculated.
‡ analysis performed on the first set of in-shoe pressure recordings which were made.
rhs - right heel-strike; lto - left toe-off; lhs - left heel-strike; rto - right toe-off.

APPENDIX D

Table D-I. Local peak pressures (kPa) (± 1sd) beneath the feet of the subjects during shod walking.

Subject	Left foot						Right foot					
	heel	mth4	mth3	mth2	mth1	hallux	hallux	mth1	mth2	mth3	mth4	heel
1	156	125	152	162	284	77	132	249	184	185	130	163
	159	119	140	140	347	93.1	109	213	192	195	117	161
	165	141	192	188	231	66.5	85.4	148	216	230	152	154
2†	219	124	181	167	149	268	273	247	143	177	170	233
	215	152	210	181	126	237	278	161	154	226	183	225
	210	112	174	185	138	263	313	281	143	159	113	220
3	162	92	150	146	150	198	96	145	140	125	98.5	156
	183	205	240	225	110	110	106	132	132	133	102	156
	156	145	186	166	134	174	87.7	111	161	172	133	150
4‡	205	118	240	310	164	175	154	164	259	191	145	144
	203	161	251	300	136	181	155	159	242	187	149	137
	205	123	229	287	178	201	147	167	262	195	154	130
5	320	205	288	262	176	149	104	221	355	348	246	191
	308	160	269	262	197	184	125	289	300	299	188	200
	375	150	251	261	256	189	92	220	346	372	253	186
6	171	171	287	266	212	119	139	157	254	231	180	166
	179	175	282	263	199	116	134	161	253	246	180	170
	179	187	292	269	202	95.9	137	230	239	210	125	148

† pressure recordings not obtained from both feet simultaneously.
‡ analysis performed on the first set of in-shoe pressure recordings which were made.

Table D-I (continued). Local peak pressures (kPa) (± 1sd) beneath the feet of the subjects during shod walking.

Subject	Left foot						Right foot					
	heel	mth4	mth3	mth2	mth1	hallux	hallux	mth1	mth2	mth3	mth4	heel
7‡	177	132	177	208	140	287	306	211	136	125	91.7	121
	174	153	188	213	146	252	328	226	141	137	87.3	129
	169	135	185	197	158	239	244	173	152	143	111	126
8†	166	98.2	242	230	224	172	231	221	260	319	154	180
	166	96.6	229	215	267	181	235	321	260	288	137	166
	169	59.4	212	204	305	203	229	310	255	298	134	189
9†	180	211	298	339	144	125	121	177	332	327	179	178
	177	171	315	283	188	132	118	160	321	318	202	188

† pressure recordings not obtained from both feet simultaneously.
‡ analysis performed on the first set of in-shoe pressure recordings which were made.

Table D-II. Local peak pressures (kPa) (± 1sd) beneath the feet of the patients during shod walking.

Patient	Left foot						Right foot					
	heel	mth4	mth3	mth2	mth1	hallux	hallux	mth1	mth2	mth3	mth4	heel
1	276	181	235	237	369	245	164	274	159	124	49.4	198
	272	246	275	299	252	236	116	213	189	138	118	185
	280	168	211	217	373	244	146	208	178	125	95	188
2	149	230	278	292	97.3	161	319	136	305	316	306	118
	109	186	219	227	161	85.9	267	173	253	284	293	124
	-	-	-	-	-	-	195	88.2	176	225	230	187
3†	154	127	175	158	256	141	-	-	-	-	-	-
	179	148	198	164	229	134	-	-	-	-	-	-
	167	170	288	276	303	143	-	-	-	-	-	-
4	289	168	210	177	269	176	27	313	169	137	76.1	223
	276	82	129	125	357	112	36.7	194	259	228	142	237
	316	98.2	154	156	332	147	47	278	213	156	128	193
5	165	142	255	376	244	180	85.6	123	225	185	131	193
	181	149	249	359	231	176	102	177	231	169	105	175
	196	177	253	354	210	164	88.3	151	239	178	116	170

† pressure recordings were not obtained from beneath the right foot.

Table D-II (continued). Local peak pressures (kPa) (± 1sd) beneath the feet of the patients during shod walking.

Patient	Left foot						Right foot					
	heel	mth4	mth3	mth2	mth1	hallux	hallux	mth1	mth2	mth3	mth4	heel
6†	-	-	-	-	-	-	193	563	180	238	202	187
	-	-	-	-	-	-	127	229	123	327	320	216
7‡	228	288	230	376	315	93	282	401	523	357	228	214
	262	288	233	352	224	71.4	175	429	505	388	479	211
	-	-	-	-	-	-	161	310	485	384	380	188

† pressure data not included from the left foot (creasing of the F-Scan insole inside the shoe made the pressure data unreliable).
‡ pressure recordings not obtained from both feet simultaneously.

APPENDIX E

Table E-I. Times of onset, maximum peak and cessation of pressure at local sites beneath the right feet of the subjects during shod walking (expressed as a percentage of the stance phase).

Subject	heel			mth4			mth3			mth2			mth1			hallux		
	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation
1	0	20.3	49.3	4.3	71.0	98.6	8.7	79.7	100	10.1	79.7	100	11.6	75.4	98.6	24.6	76.8	95.7
	0	18.2	42.4	7.6	65.2	95.5	12.1	78.8	100	12.1	78.8	98.5	10.6	74.2	98.5	25.8	78.8	95.5
	0	20.6	44.1	5.9	64.7	100	11.8	80.9	100	14.7	80.9	100	11.8	72.1	100	25.0	80.9	97.1
2	0	15.9	61.9	15.9	69.8	88.9	17.5	73.0	88.9	22.2	73.0	87.3	52.4	82.5	98.4	12.7	88.9	100
	0	17.5	57.1	14.3	71.4	90.5	14.3	73.0	90.5	17.5	74.6	88.9	60.3	84.1	98.4	12.7	90.5	100
	0	33.3	60.3	17.5	69.8	87.3	17.5	71.4	87.3	22.2	73.0	87.3	33.3	82.5	96.8	11.1	88.9	100
3	0	23.4	48.4	9.4	68.8	98.4	9.4	73.4	98.4	14.1	78.1	98.4	18.8	78.1	98.4	12.5	82.8	100
	0	25.4	46.0	9.5	71.4	98.4	11.1	71.4	98.4	15.9	76.2	98.4	17.5	76.2	98.4	11.1	82.5	100
4	1.4	16.9	50.7	18.3	71.8	97.2	19.7	74.6	97.2	23.9	81.7	94.4	33.8	78.9	94.4	23.9	85.9	98.6
	1.4	17.4	50.7	15.9	76.8	97.1	15.9	76.8	97.1	18.8	79.7	95.7	29.0	81.2	95.7	17.4	88.4	100
5	0	19.4	92.5	4.5	73.1	100	13.4	77.6	100	11.9	77.6	100	14.9	82.1	100	53.7	86.6	98.5
	0	25.4	91.0	3.0	73.1	100	11.9	77.6	100	13.4	77.6	100	13.4	77.6	100	53.7	85.1	98.5
	0	16.4	89.6	4.5	74.6	98.5	10.4	74.6	97.0	10.4	79.1	100	11.9	79.1	100	11.9	79.1	100
6	0	9.7	50.0	9.7	79.0	100	12.9	82.3	98.4	14.5	82.3	100	12.9	77.4	100	56.5	82.3	98.4
	1.5	9.1	56.1	12.1	77.3	98.5	16.7	81.8	98.5	16.7	80.3	98.5	16.7	77.3	100	53.0	84.8	98.5
7	0	18.8	65.6	10.9	70,3	95.3	10.9	71.9	96.9	14,1	78.1	98.4	17.2	82.8	98.4	34.4	85.9	100
	0	11.3	46.8	9.7	71.0	95.2	9.7	71.0	96.8	9.7	75.8	96.8	9.7	82.3	100	17.7	83.9	100
	0	19.0	58.7	11.1	73.0	93.7	11.1	73.0	95.2	11.1	76.2	96.8	11.1	79.4	98.4	22.2	82.5	98.4
8	0	23.5	58.8	11.8	72.1	100	10.3	77.9	100	13.2	77.9	100	17.6	79.4	100	10.3	88.2	100
	0	20.6	64.7	10.3	73.5	100	8.8	76.5	100	13.2	79.4	100	16.2	82.4	100	10.3	88.2	100
	0	22.4	62.7	9.0	73.1	100	7.5	77.6	100	10.4	79.1	100	13.4	80.6	100	9.0	88.1	100
9	0	21.3	74.7	6.7	76.0	97.3	8.0	78.7	100	10.7	80.0	100	12.0	77.3	98.7	29.3	85.3	96.0
	0	21.3	74.7	6.7	76.0	100	8.0	81.3	100	9.3	82.7	100	12.0	81.3	100	64.0	89.3	98.7

Table E-II. Times of onset, maximum peak and cessation of pressure at local sites beneath the left feet of the subjects during shod walking (expressed as a percentage of the stance phase).

Subject	heel			mth4			mth3			mth2			mth1			hallux		
	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation
1	0	17.6	58.8	8.8	60.3	91.2	13.2	73.5	94.1	13.2	77.9	94.1	13.2	73.5	100	30.9	70.6	97.1
	0	22.4	67.2	7.5	49.3	91.0	11.9	70.1	94.0	11.9	70.1	98.5	10.4	76.1	100	49.3	74.6	98.5
	0	19.1	50.0	5.9	66.2	94.1	10.3	75.0	95.6	10.3	79.4	98.5	10.3	75.0	100	45.6	75.0	98.5
2	0	25.8	69.7	22.7	74.2	90.9	22.7	75.8	92.4	28.8	77.3	92.4	39.4	78.8	97.0	15.2	87.9	100
	0	18.5	52.3	18.5	72.3	90.8	21.5	78.5	92.3	27.7	78.5	92.3	61.5	83.1	95.4	13.8	87.7	100
	0	22.7	63.6	24.2	72.7	89.4	30.3	78.8	90.9	34.8	78.8	92.4	43.9	78.8	97.0	15.2	86.4	100
3	0	18.8	48.4	3.1	70.3	100	7.8	78.1	100	14.1	78.1	100	17.2	78.1	100	1.6	87.5	100
	0	19.1	47.1	4.4	75.0	100	4.4	75.0	100	14.7	76.5	100	19.1	76.5	97.1	1.5	85.3	100
	0	20.0	44.6	0	70.8	100	4.6	73.8	100	15.4	78.5	100	18.5	78.5	100	3.1	87.7	100
4	0	26.0	71.2	20.5	76.7	98.6	23.3	80.8	98.6	23.3	80.8	98.6	37.0	80.8	98.6	17.8	87.7	100
	0	24.3	64.3	20.0	77.1	97.1	21.4	78.6	98.6	24.3	80.0	98.6	31.4	81.4	97.1	17.1	88.6	100
	0	20.5	67.1	20.5	74.0	95.9	20.5	78.1	97.3	20.5	80.8	97.3	27.4	78.1	95.9	16.4	87.7	98.6
5	0	16.7	87.9	12.1	74.2	100	12.1	77.3	100	13.6	77.3	100	24.2	81.8	100	18.2	86.4	100
	1.5	23.5	88.2	14.7	70.6	100	16.2	75.0	100	19.1	75.0	100	35.3	79.4	97.1	16.2	82.4	100
	0	20.6	80.9	16.2	72.1	100	16.2	75.0	100	19.1	75.0	100	35.3	75.0	100	19.1	83.8	100
6	0	8.1	61.3	21.0	85.5	98.4	21.0	85.5	100	21.0	80.6	100	16.1	77.4	100	59.7	90.3	100
	0	9.2	60.0	21.5	83.1	96.9	21.5	83.1	100	21.5	83.1	100	16.9	78.5	100	58.5	90.8	100
	1.5	9.2	50.8	20.0	81.5	100	20.0	81.5	100	16.9	78.5	100	12.3	78.5	100	61.5	90.8	100
7	0	15.9	66.7	15.9	73.0	96.8	15.9	79.4	96.8	15.9	81.0	96.8	14.3	84.1	98.4	23.8	87.3	100
	0	15.2	86.4	18.2	77.3	97.0	18.2	80.3	97.0	19.7	80.3	97.0	43.9	84.8	97.0	48.5	87.9	100
	0	12.1	84.8	15.2	77.3	95.5	15.2	80.3	97.0	15.2	80.3	100	12.1	83.3	100	25.8	86.4	100
8	0	13.5	68.9	14.9	74.3	95.9	13.5	78.4	98.6	17.6	81.1	98.6	17.6	77.0	98.6	16.2	87.8	100
	0	16.4	69.9	13.7	71.2	97.3	12.3	78.1	100	13.7	82.2	98.6	20.5	78.1	100	16.4	82.2	100
9	0	25.0	88.8	12.5	78.8	98.8	11.3	81.3	100	8.8	81.3	100	12.5	80.0	100	21.3	90.0	98.8
	0	25.9	90.1	13.6	76.5	97.5	9.9	81.5	100	11.1	84.0	100	14.8	81.5	100	19.8	88.9	98.8

Table E-III. Times of onset, maximum peak and cessation of pressure at local sites beneath the right feet of the patients during shod walking (expressed as a percentage of the stance phase).

Patient	heel			mth4			mth3			mth2			mth1			hallux		
	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation
1	1.5	33.8	77.9	20.6	76.5	85.3	20.6	76.5	91.2	20.6	76.5	94.1	20.6	83.8	98.5	54.4	91.2	98.5
	1.4	23.6	73.6	19.4	76.4	86.1	19.4	76.4	90.3	19.4	80.6	94.4	23.6	80.6	98.6	56.9	91.7	98.6
	0	27.7	73.8	21.5	75.4	86.2	18.5	75.4	90.8	18.5	80.0	95.4	21.5	80.0	100	58.5	90.8	98.5
2	0	11.6	41.9	11.6	62.8	89.5	14.0	70.9	93.0	22.1	76.7	93.0	23.3	39.5	87.2	31.4	88.4	100
	0	24.7	56.2	10.1	70.8	92.1	13.5	76.4	92.1	27.0	82.0	91.0	24.7	51.7	85.4	46.1	89.9	100
	0	26.6	74.5	7.4	78.7	90.4	10.6	80.9	90.4	59.6	84.0	92.6	64.9	78.7	88.3	70.2	90.4	98.9
3[a]	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
4	1.4	29.2	69.4	5.6	66.7	94.4	6.9	70.8	100	8.3	70.8	98.6	11.1	77.8	98.6	81.9	88.9	93.1
	1.2	14.8	60.5	3.7	71.6	97.5	7.4	75.3	100	8.6	74.1	98.8	11.1	71.6	97.5	77.8	87.7	92.6
	0	13.8	63.8	5.0	61.3	96.3	7.5	71.3	100	8.8	61.3	100	12.5	75.0	100	71.3	88.8	93.8
5	0	20.0	66.2	10.8	78.5	96.9	6.2	80.0	100	9.2	80.0	100	18.5	75.4	96.9	55.4	87.7	100
	0	27.9	66.2	8.8	76.5	92.6	5.9	79.4	100	7.4	79.4	98.5	11.8	79.4	97.1	50.0	83.8	97.1
	0	21.5	64.6	10.8	76.9	96.9	7.7	81.5	100	9.2	80.0	100	15.4	76.9	98.5	44.6	86.2	98.5
6	0	14.3	67.3	3.1	32.7	92.9	7.1	75.5	92.9	7.1	78.6	92.9	13.3	78.6	99.0	45.9	84.7	94.9
7	7.7	21.8	59.0	21.8	73.1	88.5	17.9	78.2	96.2	21.8	73.1	97.4	25.6	61.5	97.4	42.3	87.2	98.7
	3.3	13.2	45.1	12.1	68.1	91.2	12.1	75.8	100	13.2	72.5	97.8	16.5	42.9	95.6	20.9	61.5	100

[a] pressure recordings were not obtained from beneath the right foot.

Table E-IV. Times of onset, maximum peak and cessation of pressure at local sites beneath the left feet of the patients during shod walking (expressed as a percentage of the stance phase).

Patient	heel			mth4			mth3			mth2			mth1			hallux		
	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation	onset	peak	cessation
1	0	23.2	68.1	23.2	78.3	91.3	47.8	82.6	97.1	55.1	82.6	97.1	55.1	79.7	100	60.9	91.3	100
	0	17.9	65.7	47.8	79.1	97.0	47.8	83.6	98.5	50.7	83.6	98.5	56.7	83.6	98.5	56.7	88.1	100
2	0	8.0	48.3	12.6	75.9	94.3	12.6	75.9	96.6	16.1	81.6	96.6	29.9	75.9	90.8	64.4	89.7	98.9
	0	11.8	64.7	17.6	75.3	90.6	17.6	75.3	100	20.0	81.2	100	28.2	81.2	98.8	49.4	92.9	97.6
3	0	22.7	65.2	15.2	72.7	90.9	13.6	75.8	93.9	16.7	78.8	97.0	16.7	83.3	98.5	16.7	90.9	100
	1.4	30.6	72.2	16.7	76.4	94.4	13.9	76.4	95.8	16.7	80.6	95.8	50.0	86.1	97.2	26.4	91.7	100
	0	18.4	60.5	10.5	68.4	90.8	10.5	75.0	97.4	10.5	75.0	97.4	13.2	81.6	100	25.0	88.2	100
4	0	19.5	63.6	9.1	76.6	98.7	9.1	76.6	100	9.1	80.5	100	19.5	66.2	96.1	11.7	88.3	98.7
	0	26.3	76.3	13.8	81.3	96.3	13.8	81.3	98.8	22.5	81.3	96.3	20.0	77.5	96.3	16.3	85.0	97.5
5	0	19.7	53.0	9.1	74.2	97.0	6.1	78.8	100	6.1	78.8	100	7.6	74.2	100	37.9	83.3	100
	0	16.7	63.6	9.1	74.2	93.9	6.1	78.8	100	6.1	78.8	100	7.6	74.2	98.5	47.0	86.4	98.5
	1.5	24.6	63.1	10.8	75.4	98.5	6.2	78.5	100	6.2	78.5	100	9.2	78.5	100	50.8	84.6	100
6[a]	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
7	1.2	16.9	65.1	7.2	72.3	95.2	8.4	78.3	100	12.0	75.9	100	15.7	72.3	100	51.8	85.5	96.4
	0	20.5	62.7	8.4	78.3	97.6	12.0	79.5	100	18.1	85.5	97.6	18.1	73.5	100	54.2	83.1	96.4

[a] creasing of the F-Scan insole inside the shoe made the data from the left foot unreliable.

APPENDIX F

coding system for the shear records

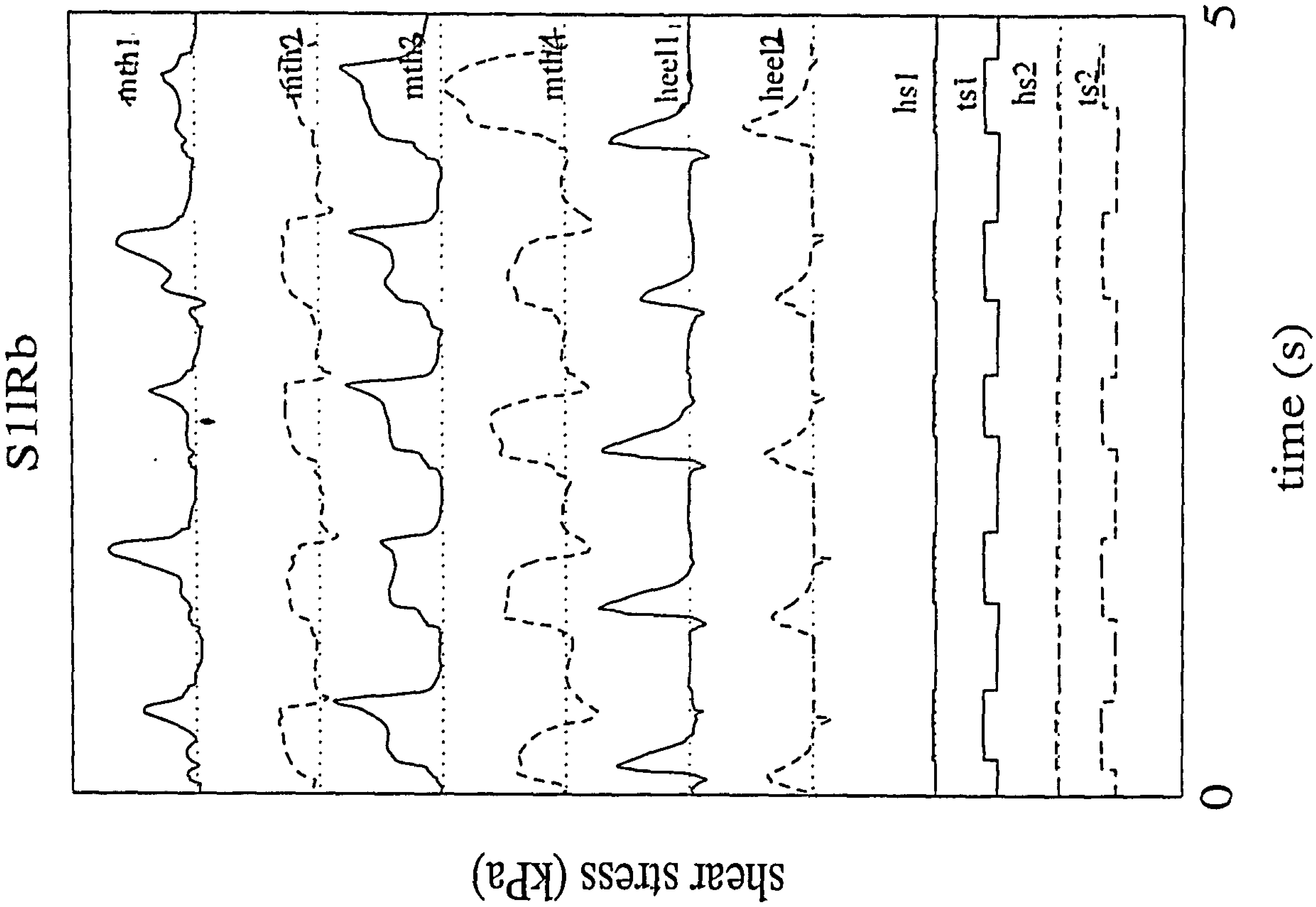
The shear records in this Appendix are titled using a simple five character code:

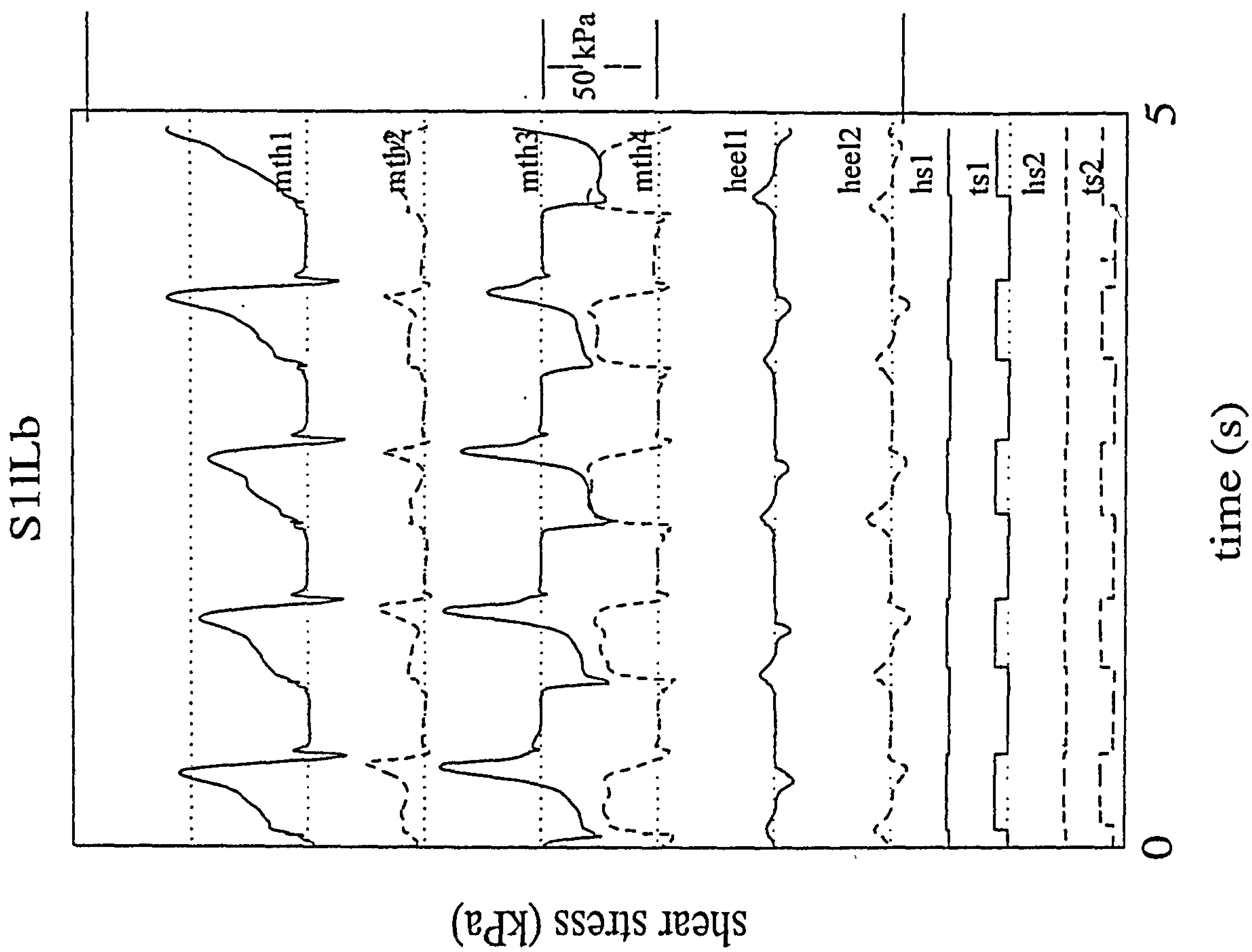
- *S* refers to a Subject shear record;
- a number refers to the particular subject (see table 5.1);
- *l* or *t* refers to a record of longitudinal or transverse shear, respectively;
- *R* or *L* refers to the Right or Left foot, respectively;

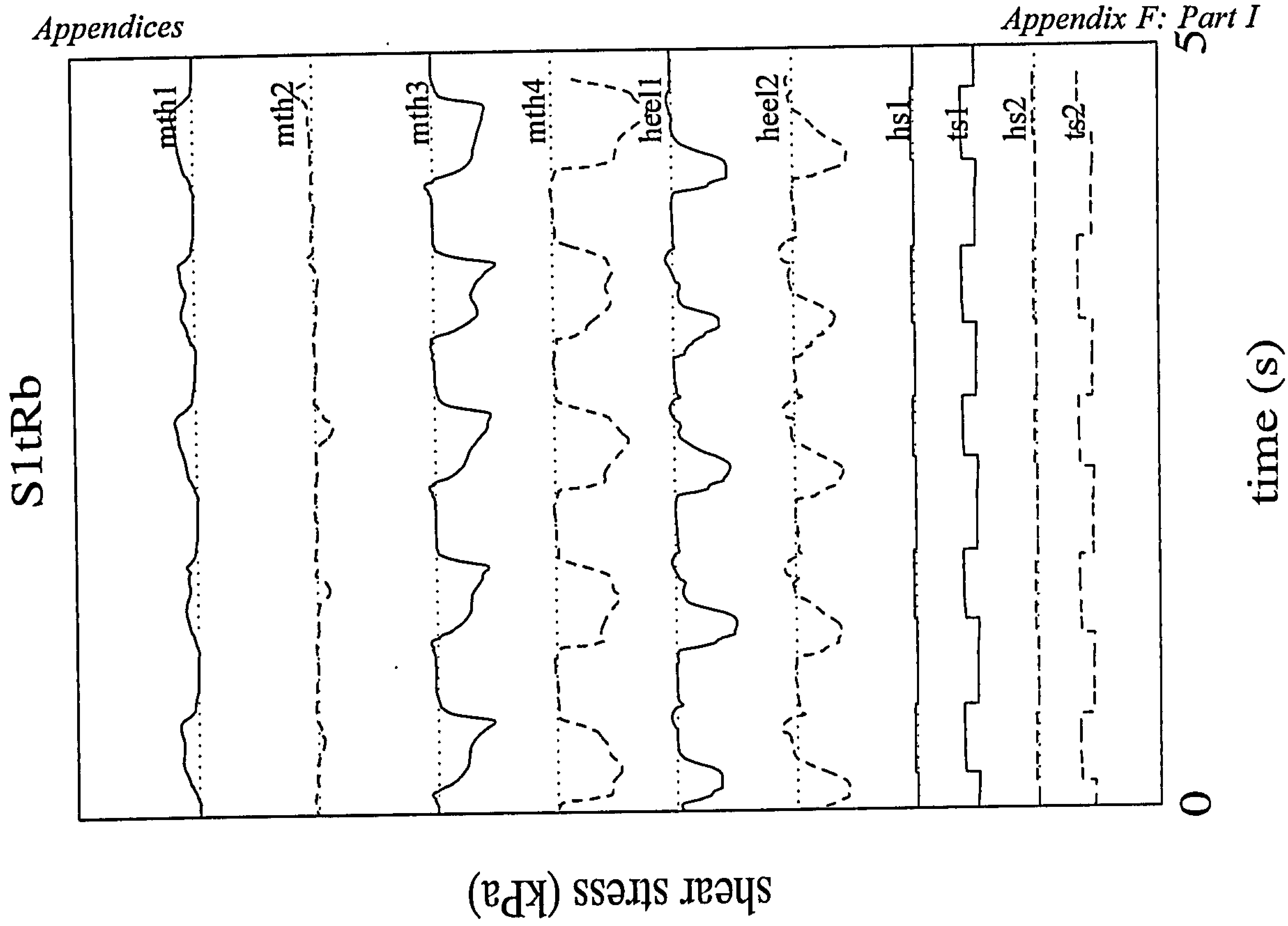
and

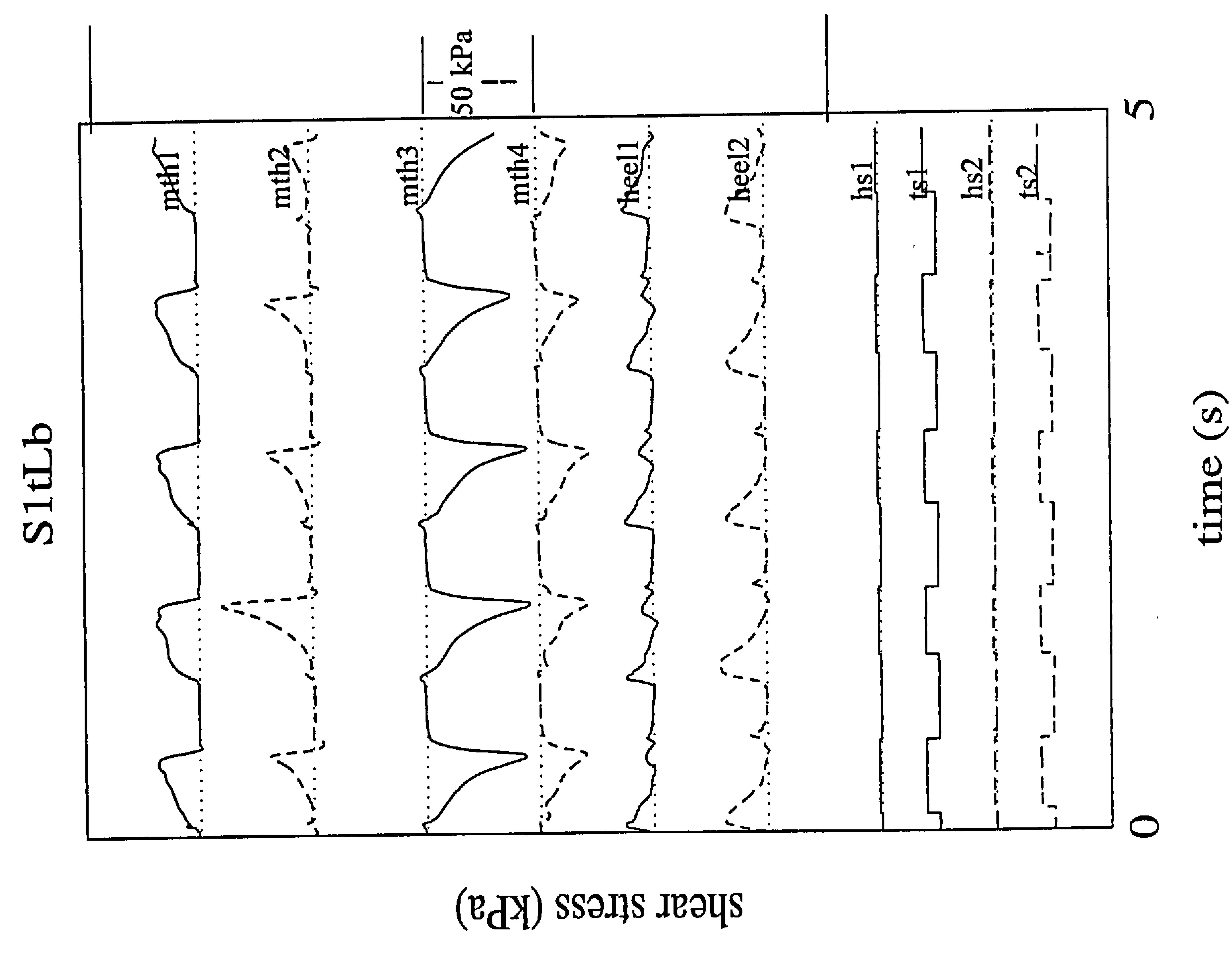
- *s* or *b* refers to the condition under which the recording was made, either with the individual wearing or not wearing nylon hold-ups, respectively, while in the extra-depth stock orthopaedic shoes ie. either wearing socks or barefoot.

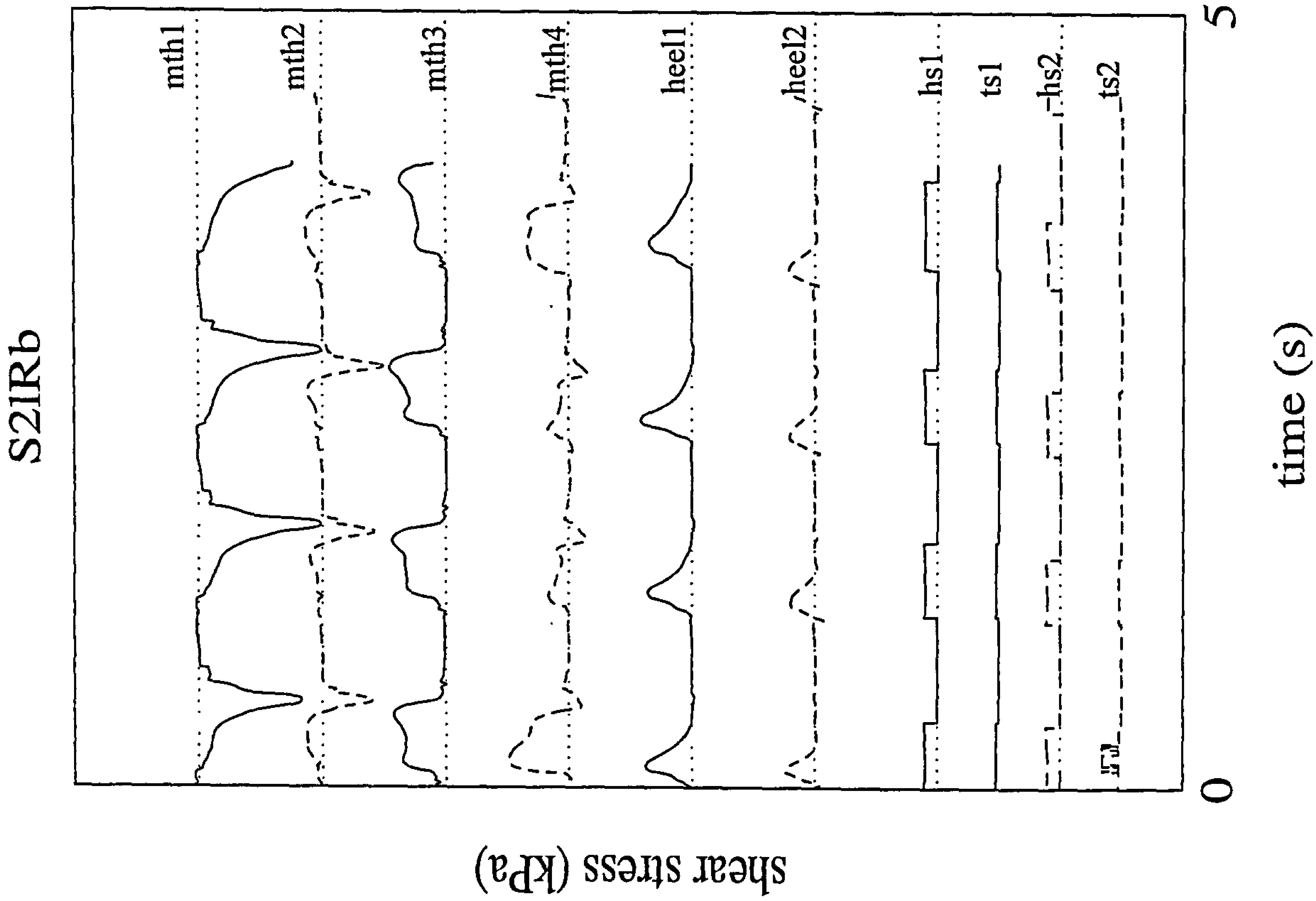
The shear records of repeat tests are referred to by the addition of the letter *R* to the end of the code described above, eg. *S9lRs-R*.

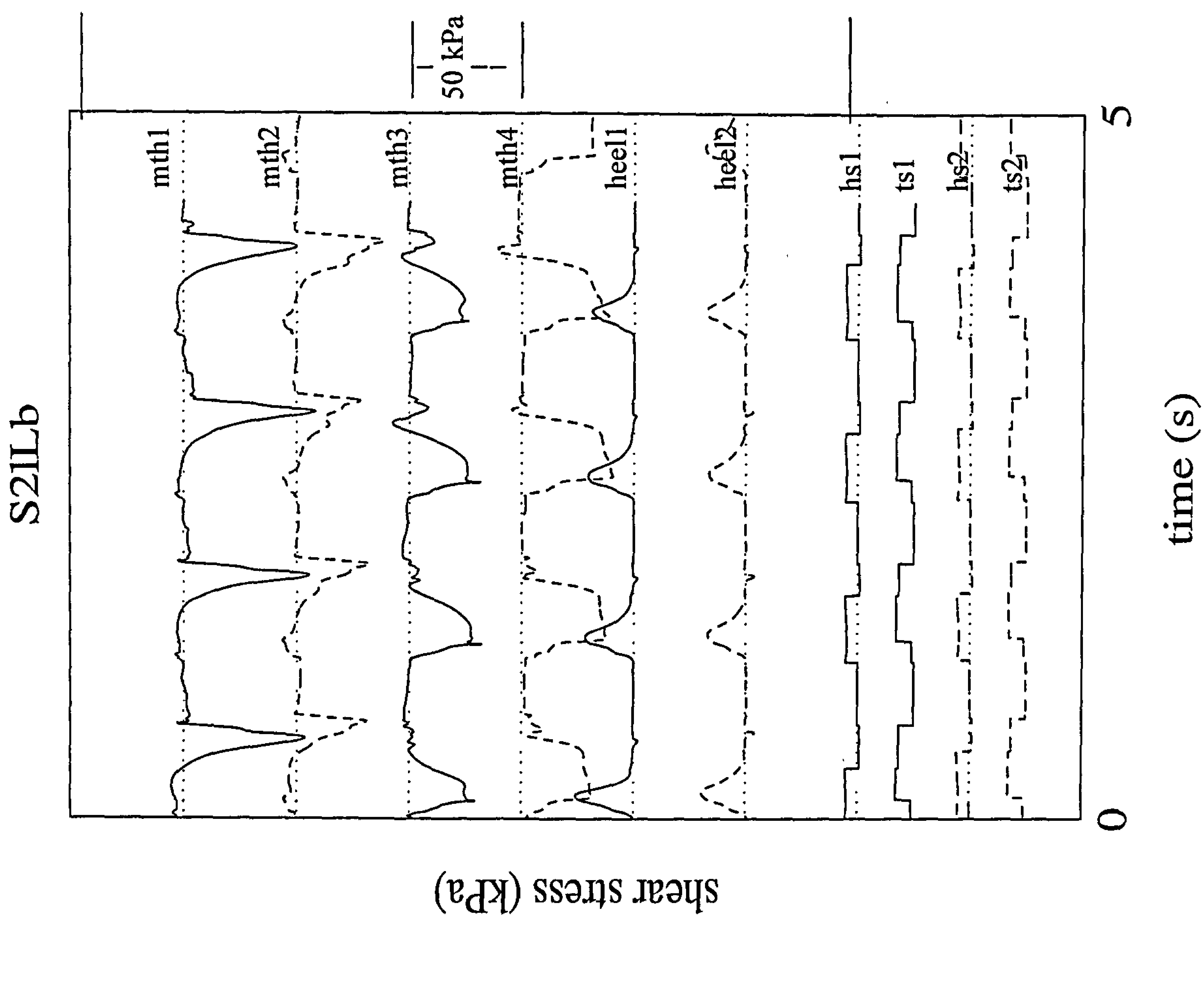

S11Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

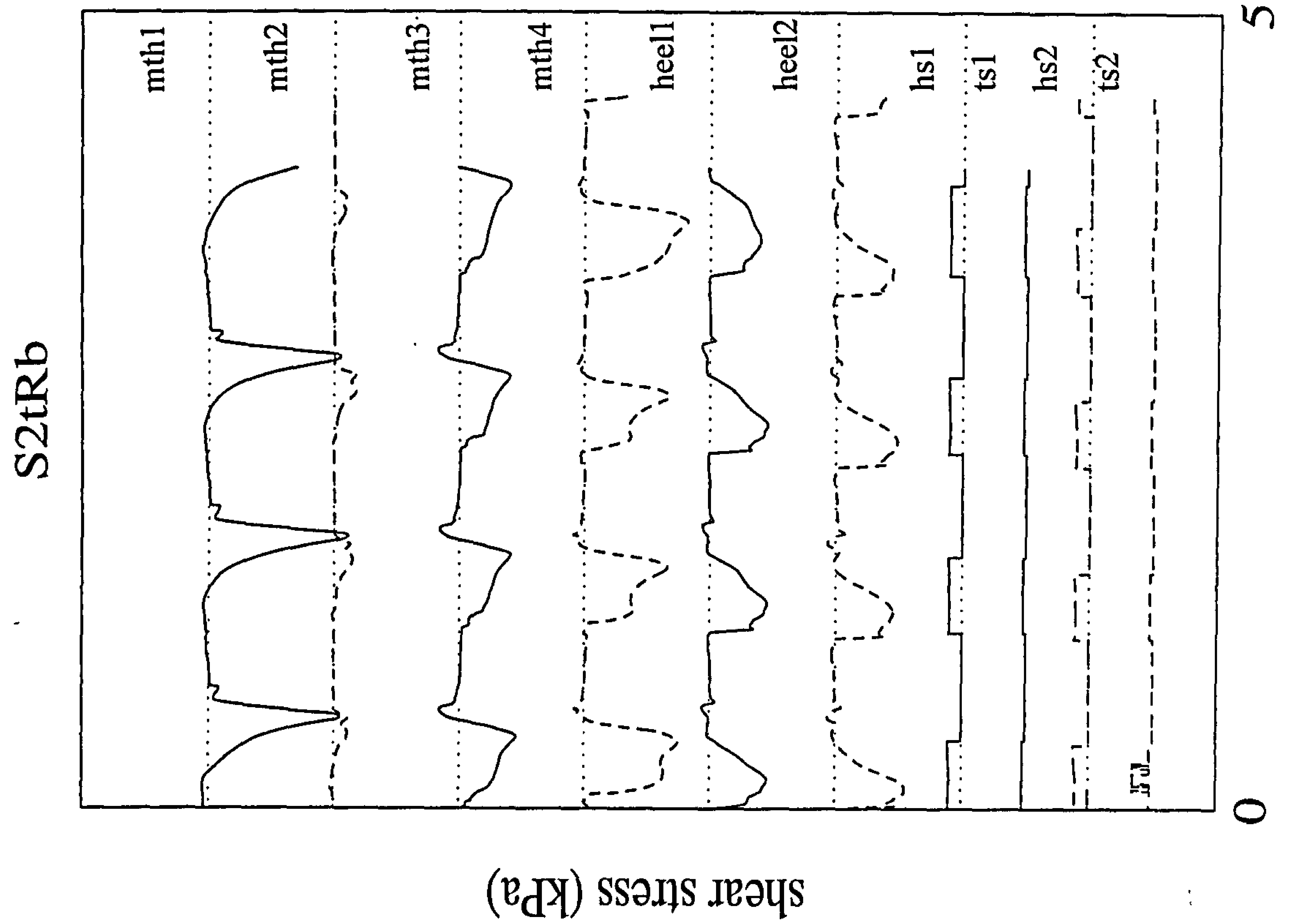

S11Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

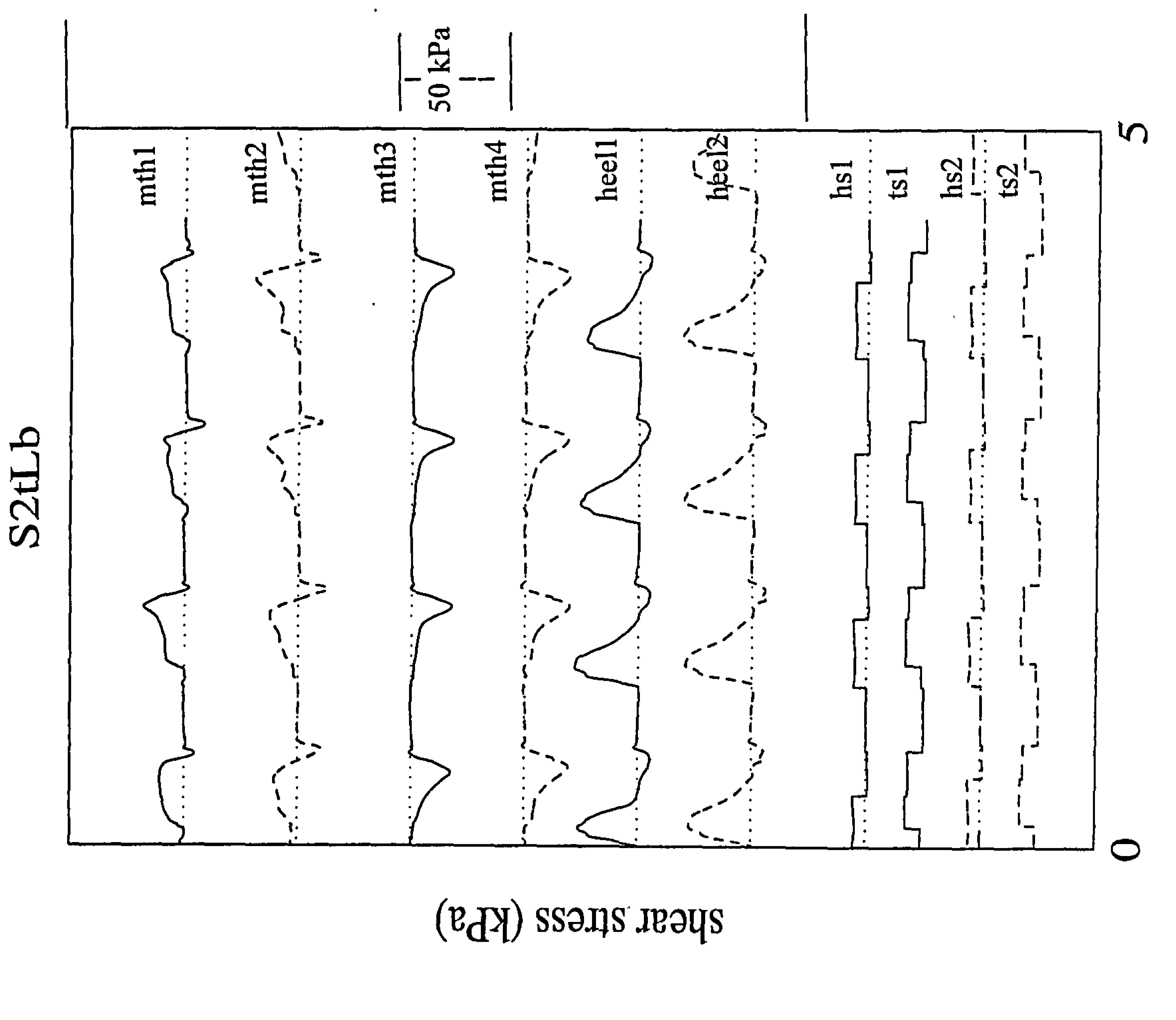

S1tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

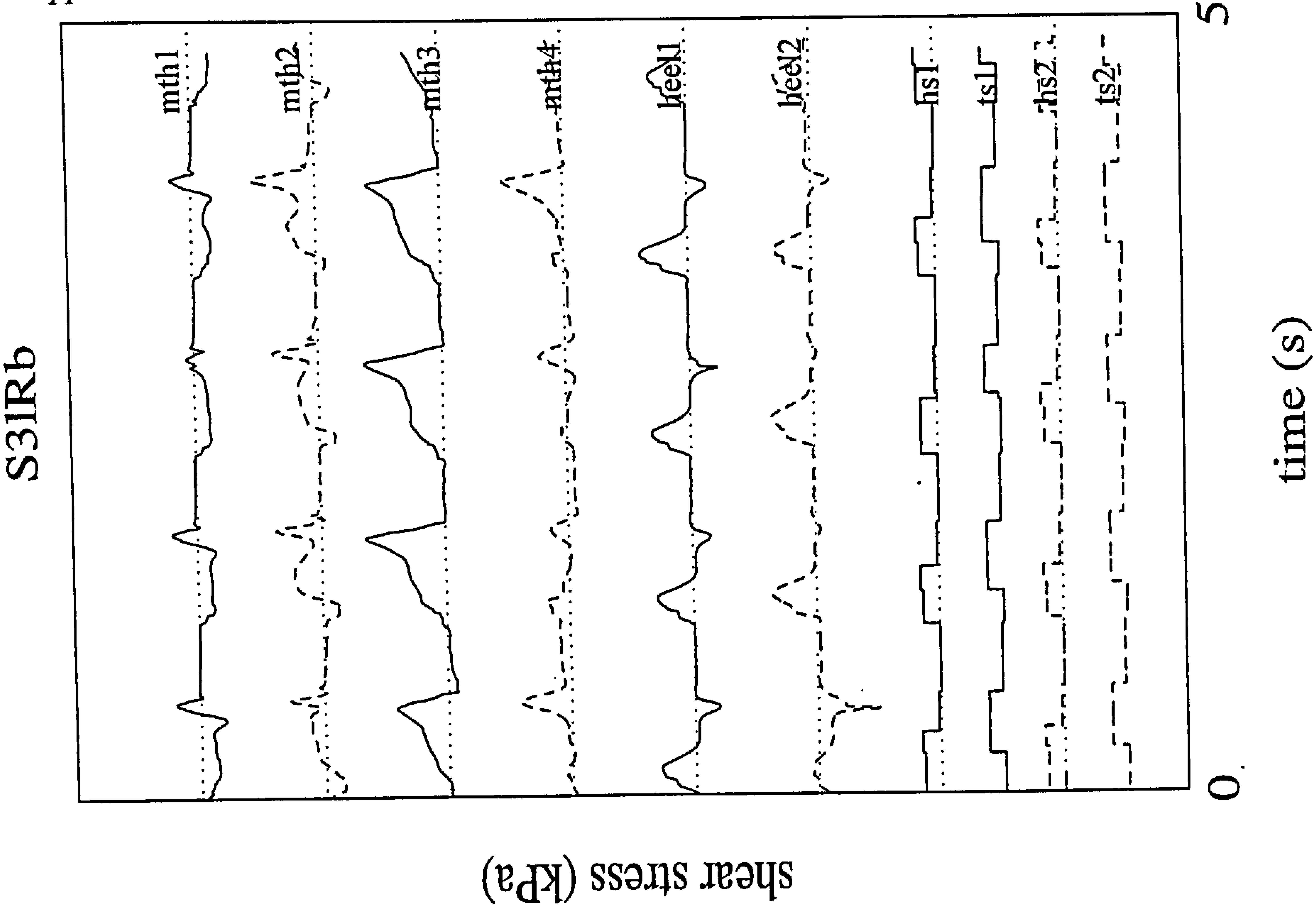

S1tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

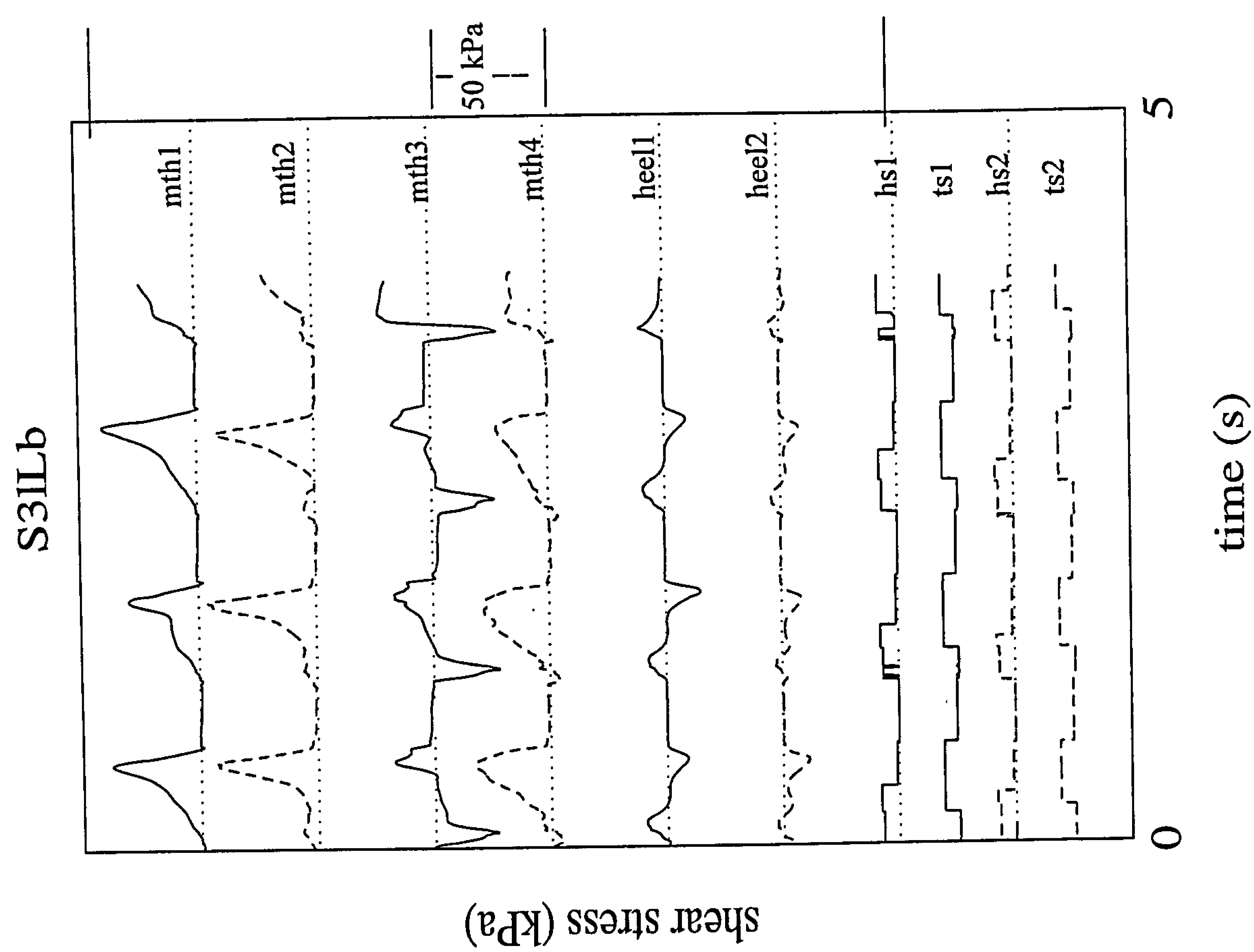

S21Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

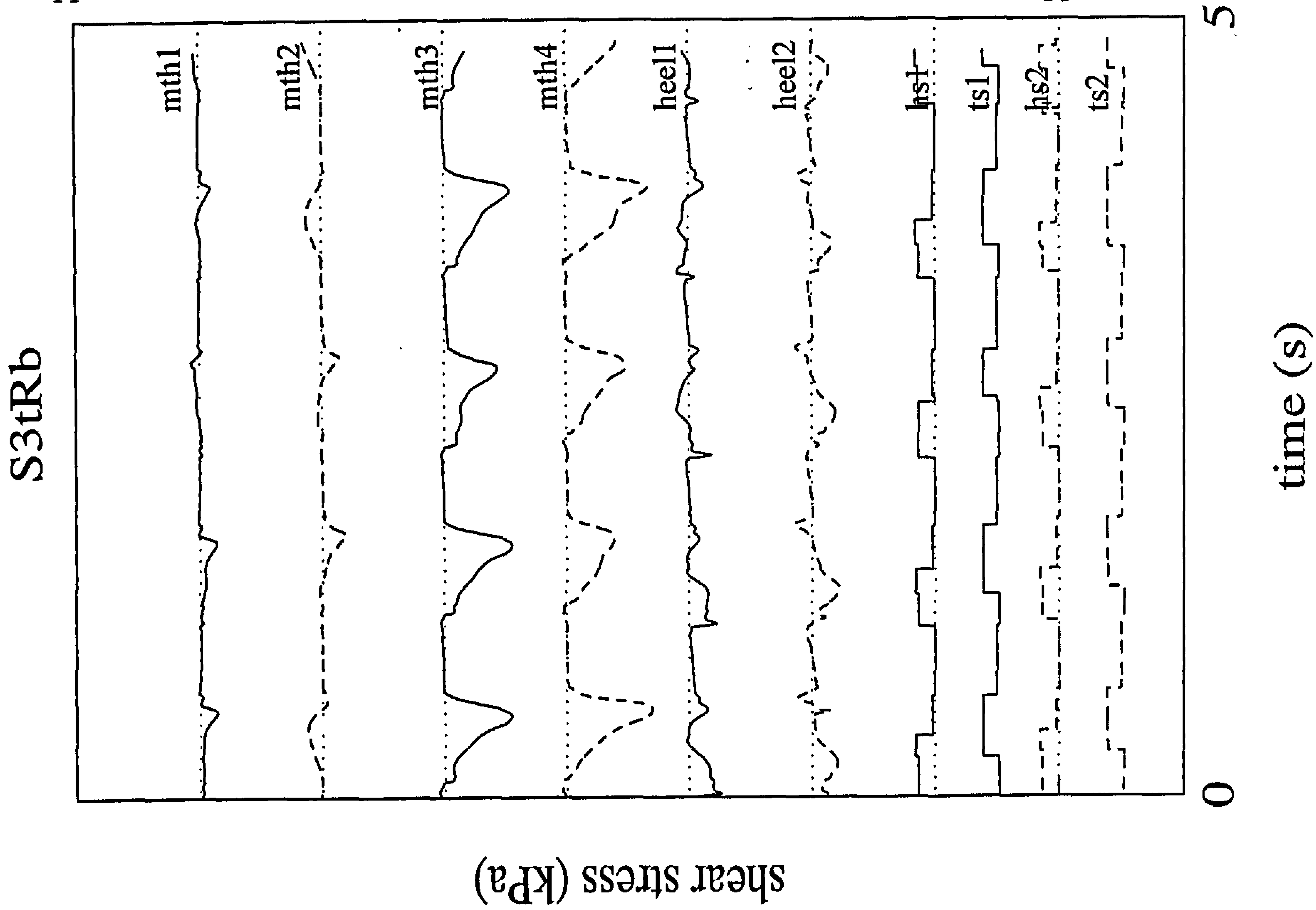

S21Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

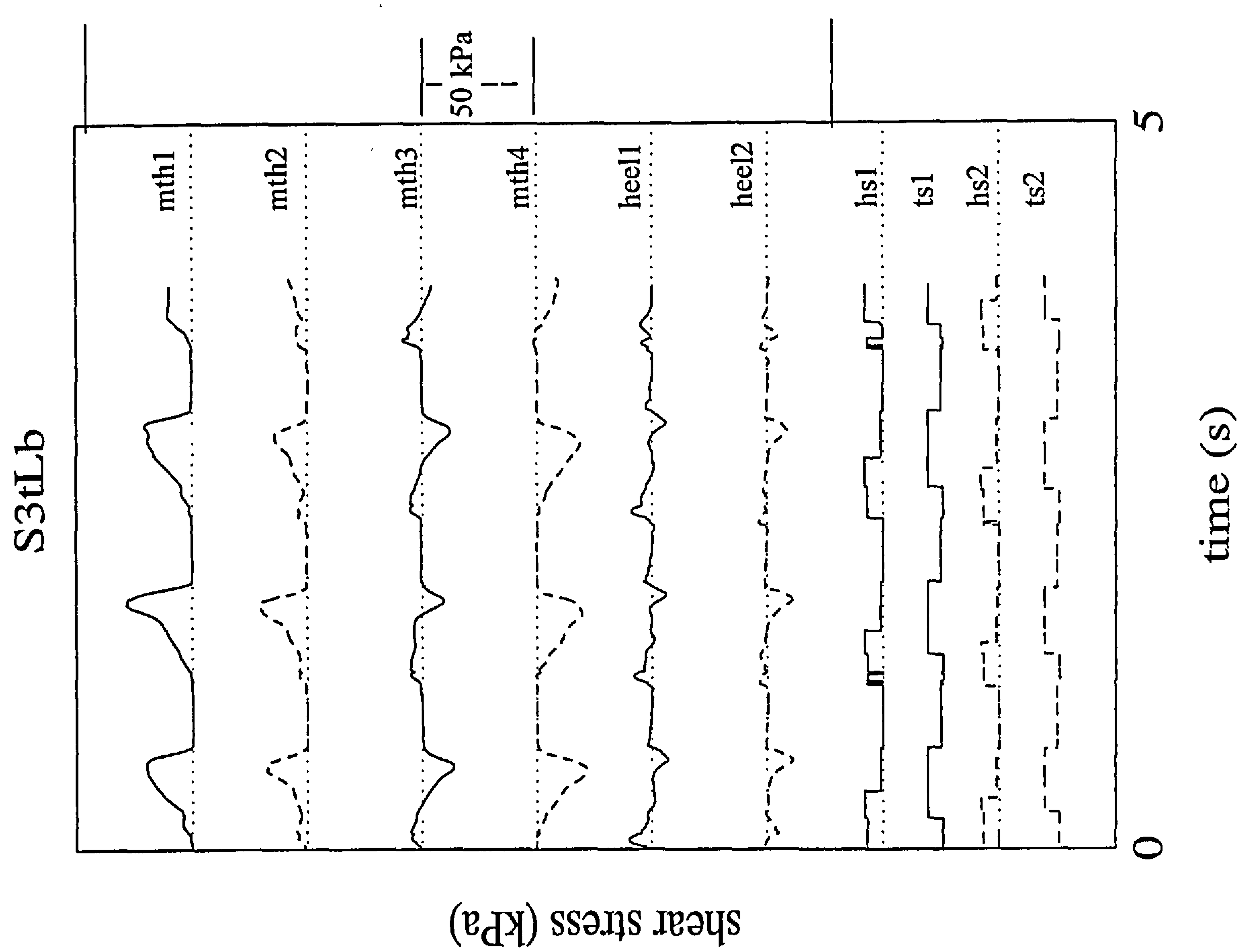

S2tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

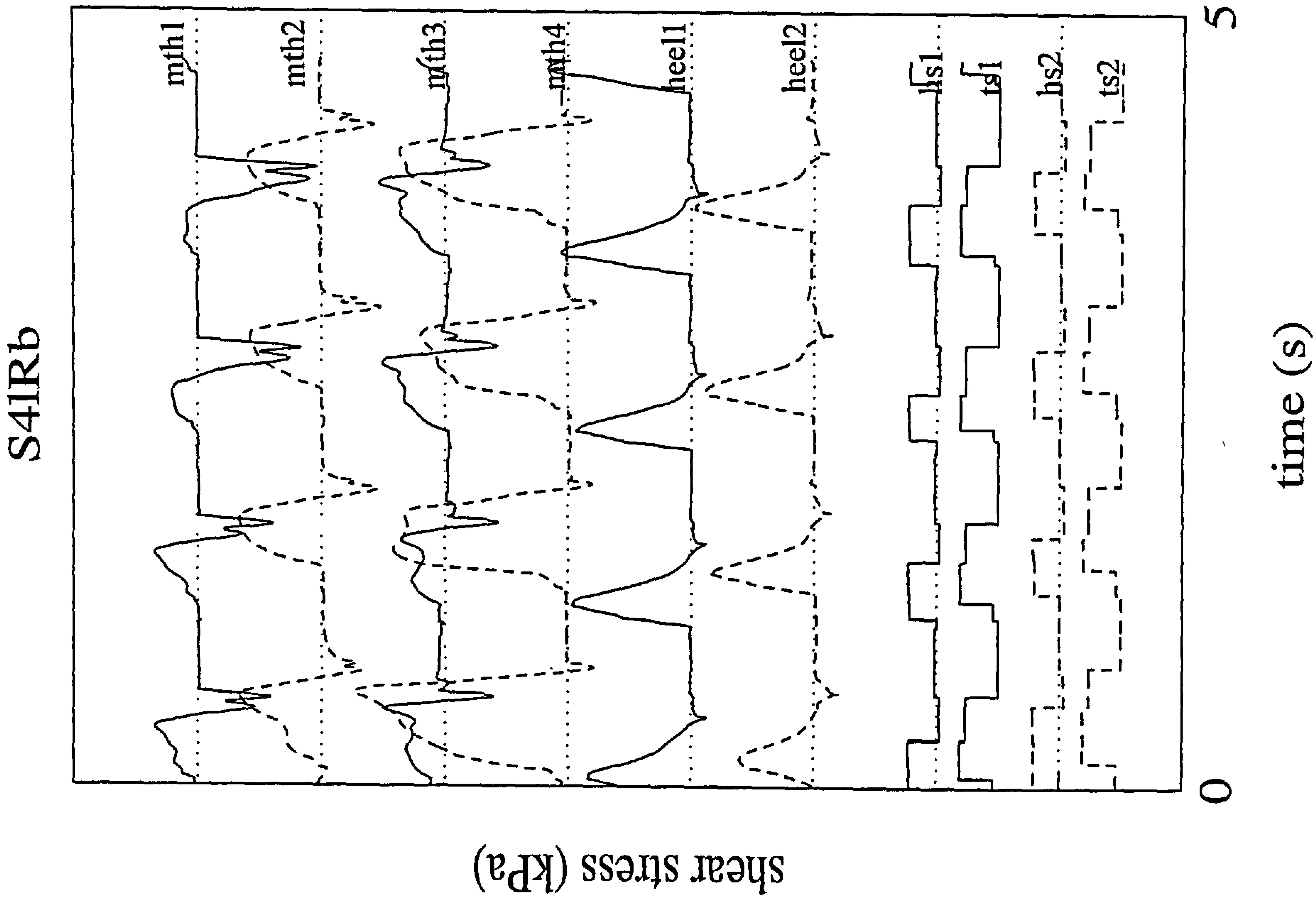

S2tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

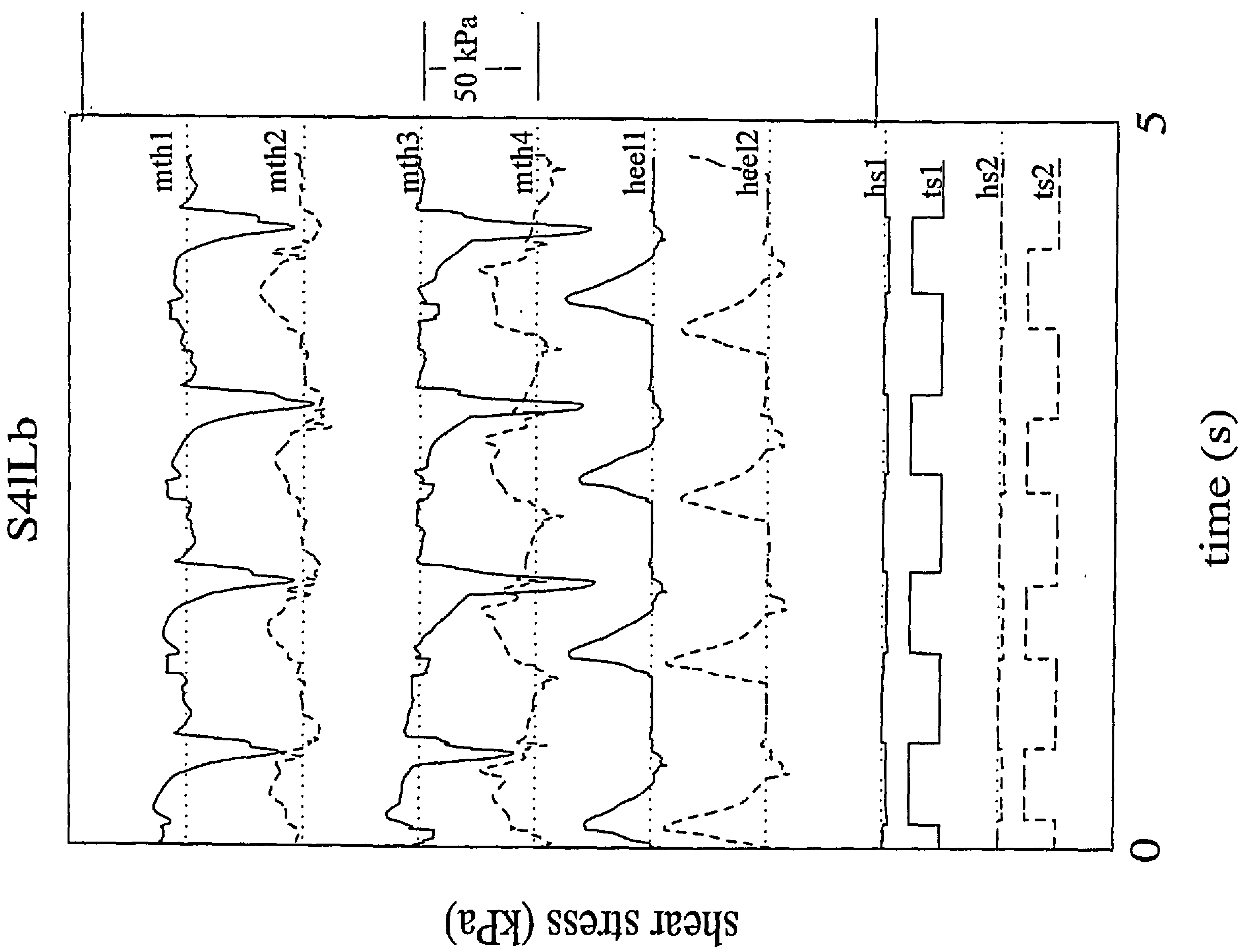

S31Rb
shear stress (kPa)
time (s)
0
5
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2

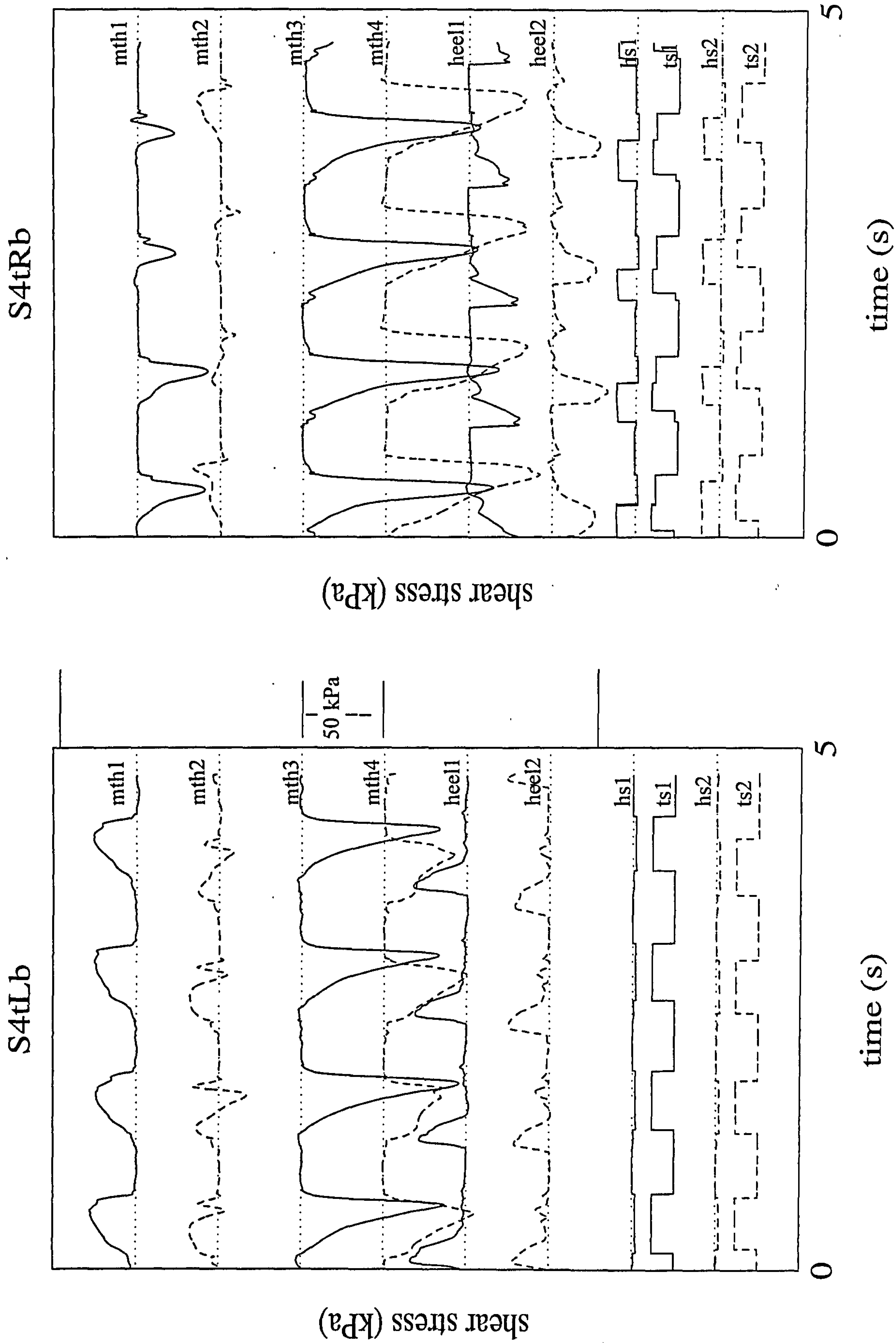

S31Lb
shear stress (kPa)
time (s)
0
5
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2

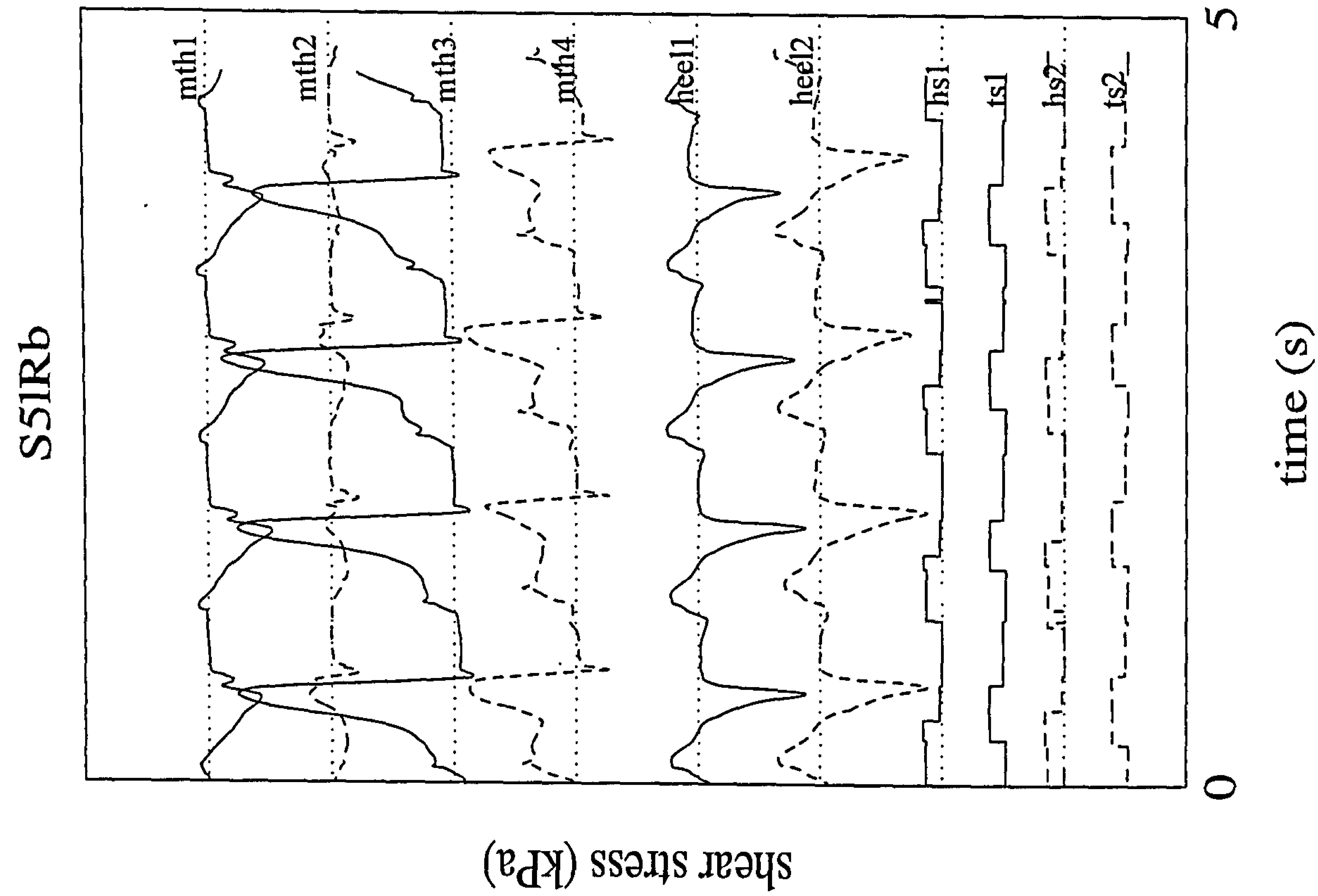

S3tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

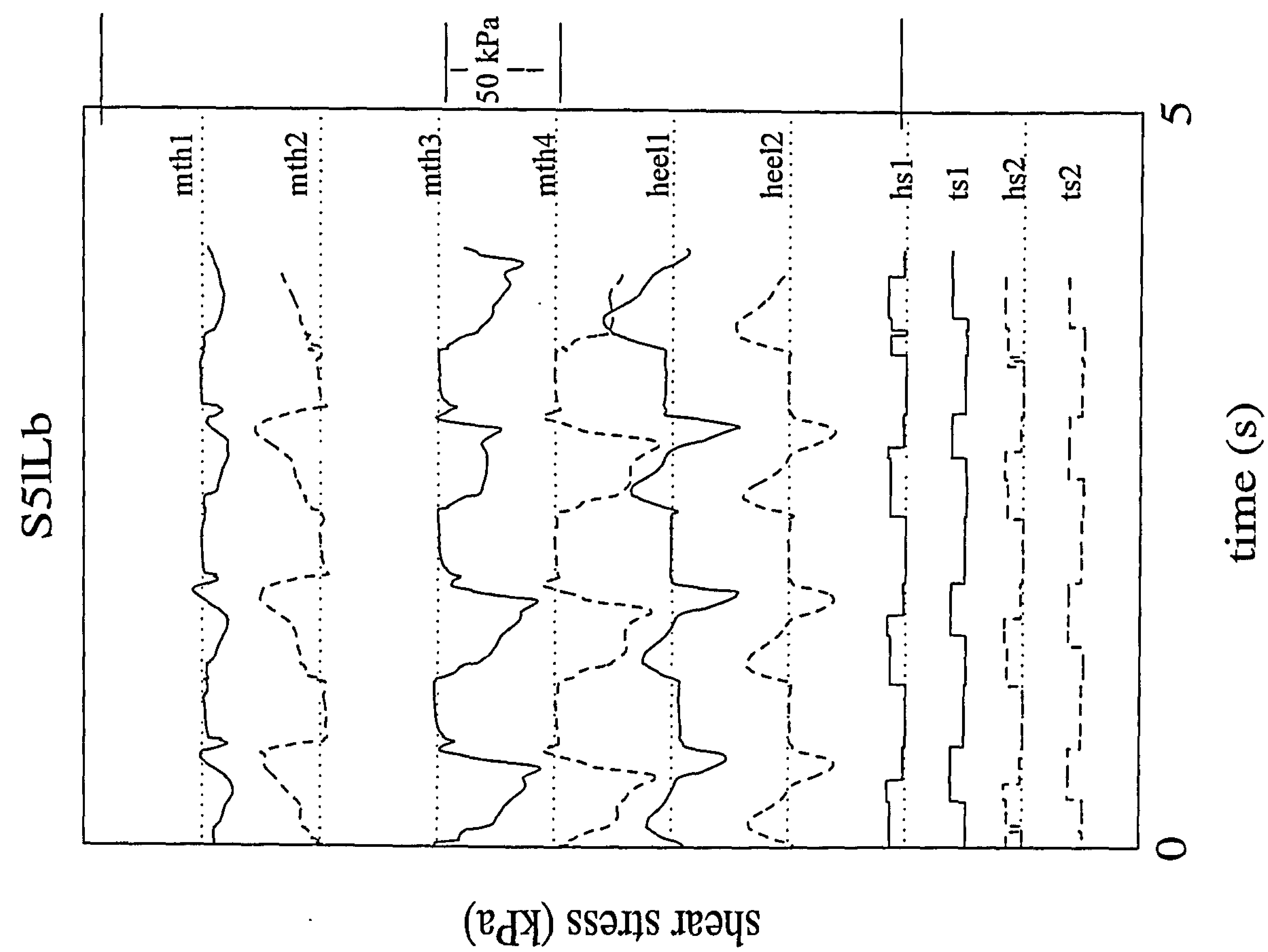

S3tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

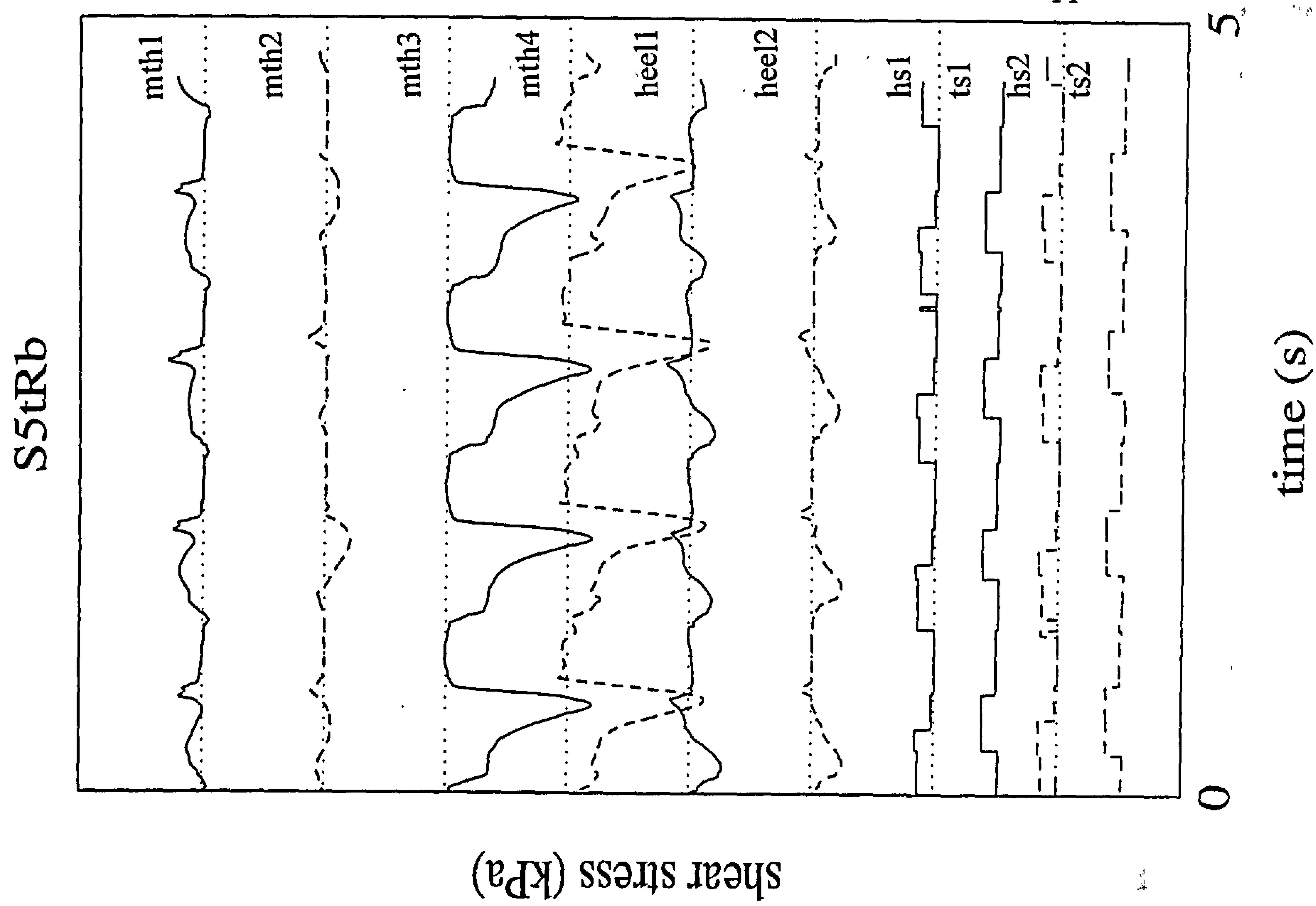

S41Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

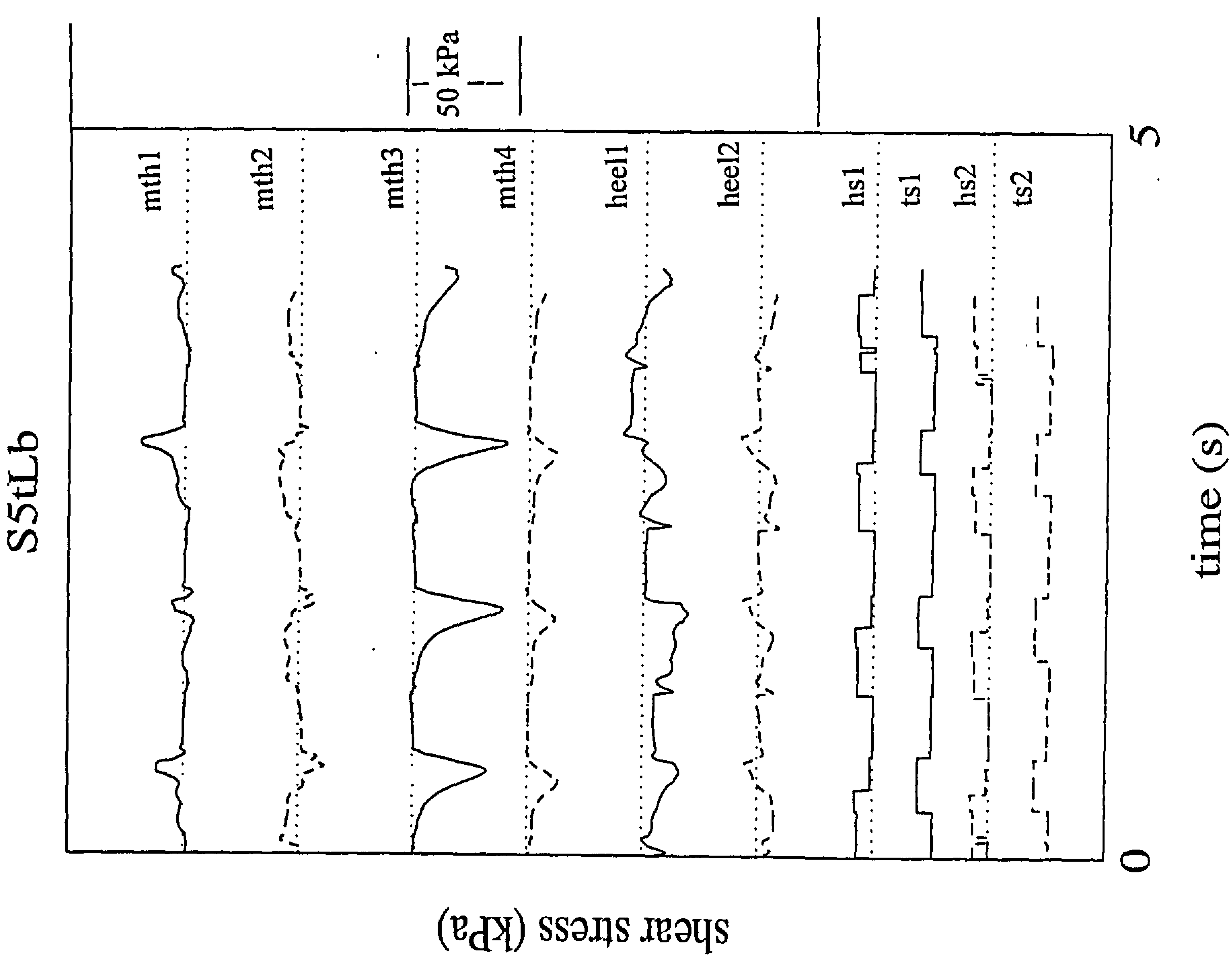

S41Lb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
time (s)
0
5

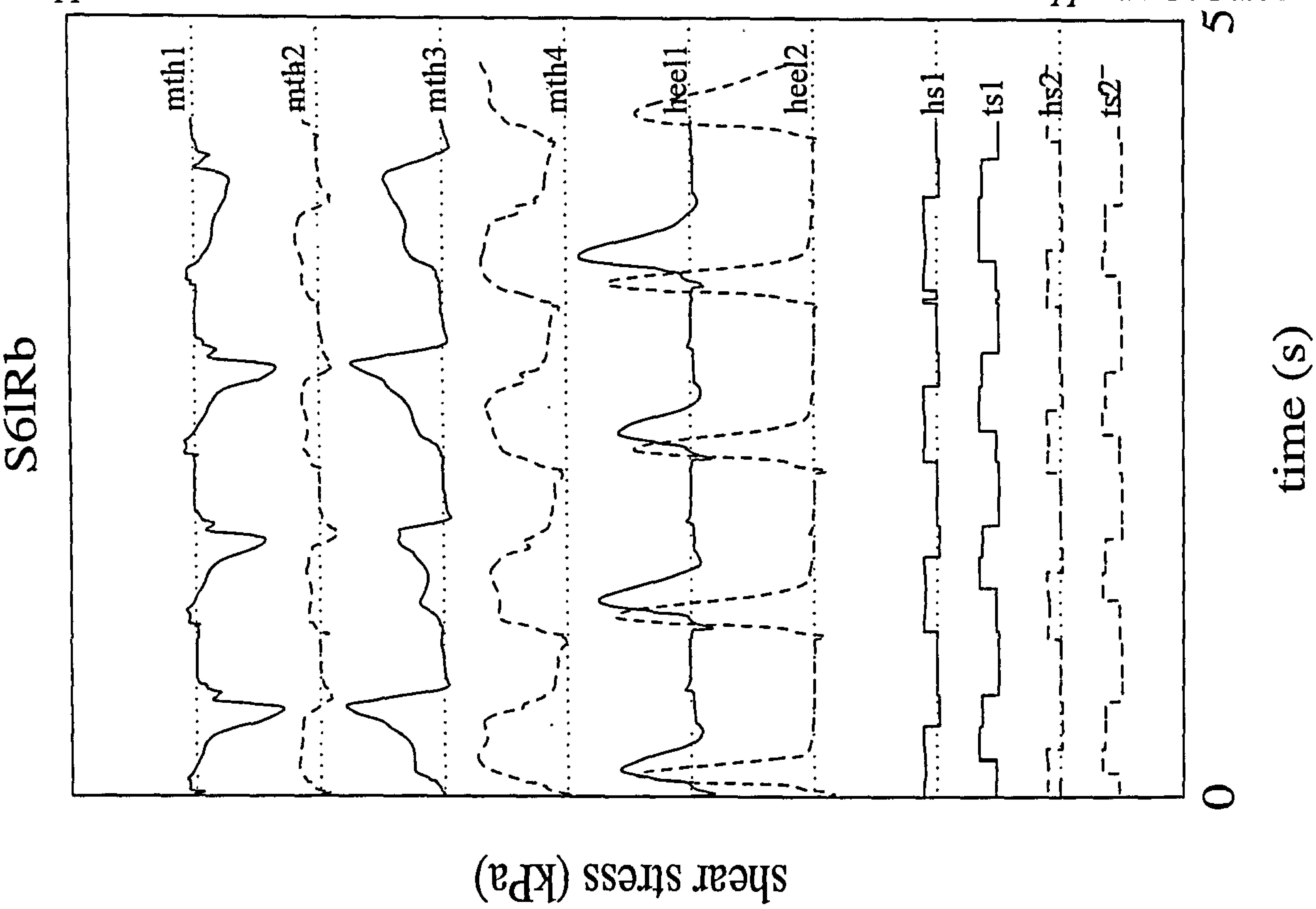

S4tRb
shear stress (kPa)
time (s)
0
5
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
S4tLb
50 kPa
shear stress (kPa)
time (s)
0
5
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2

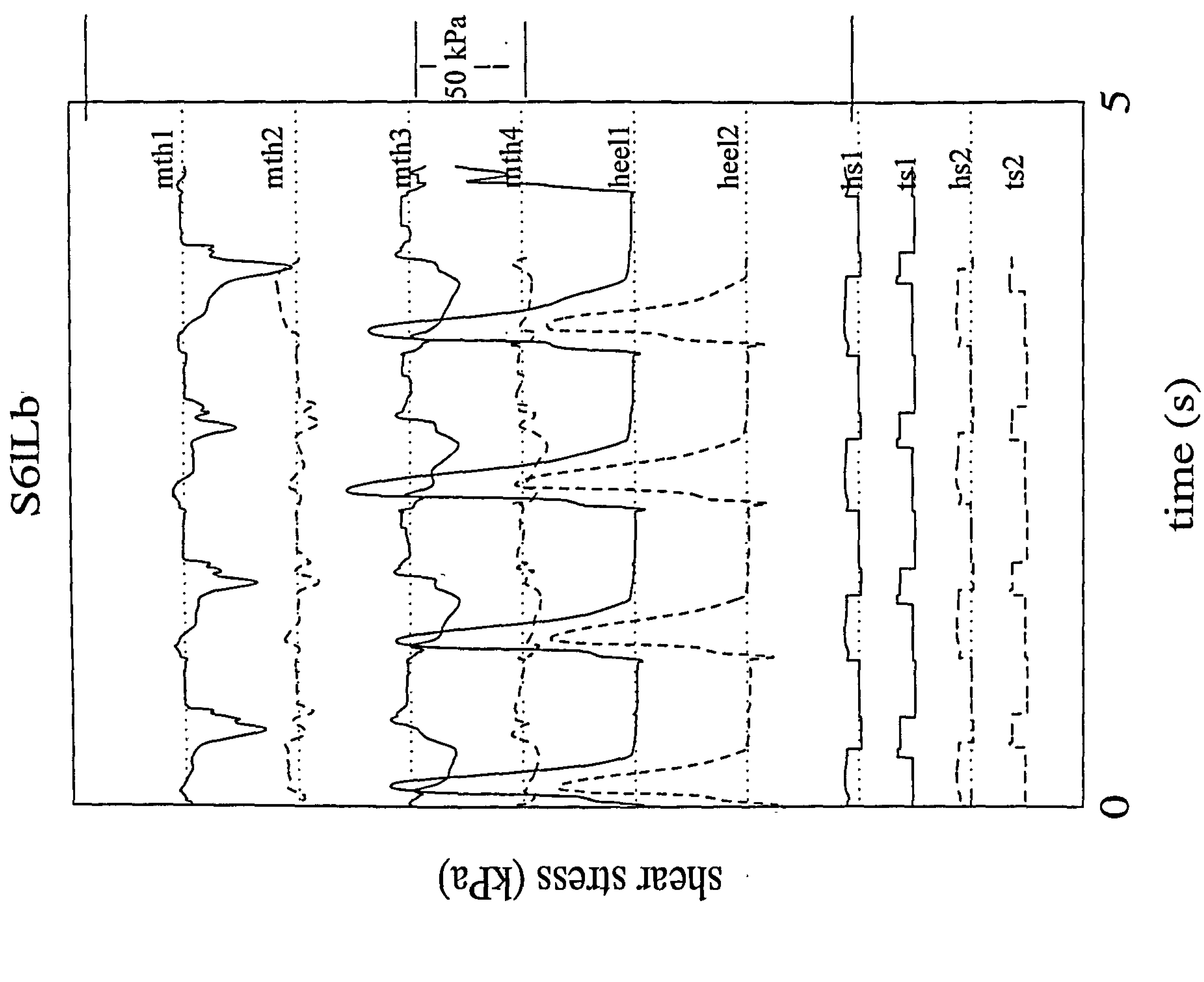

S51Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

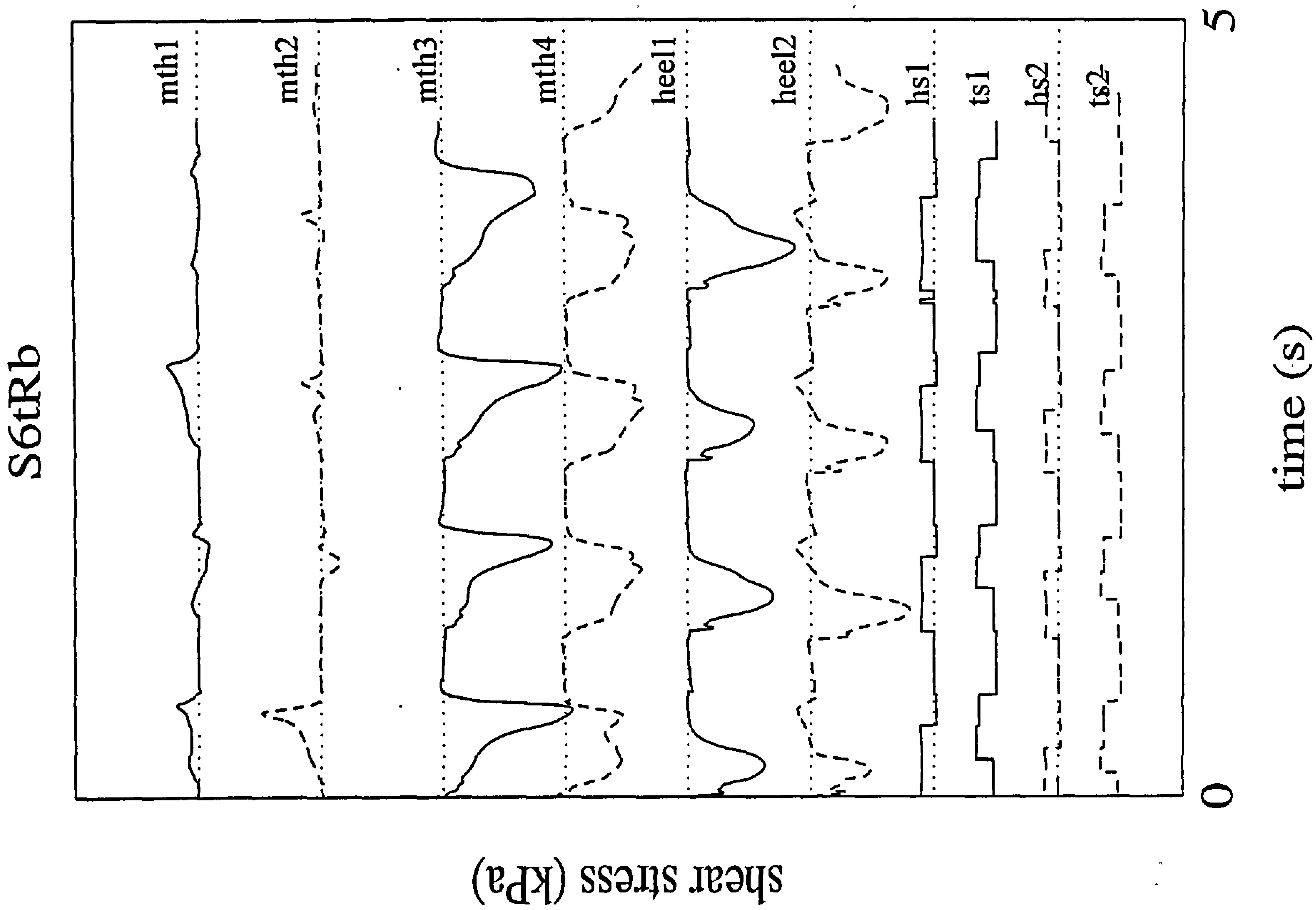

S51Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

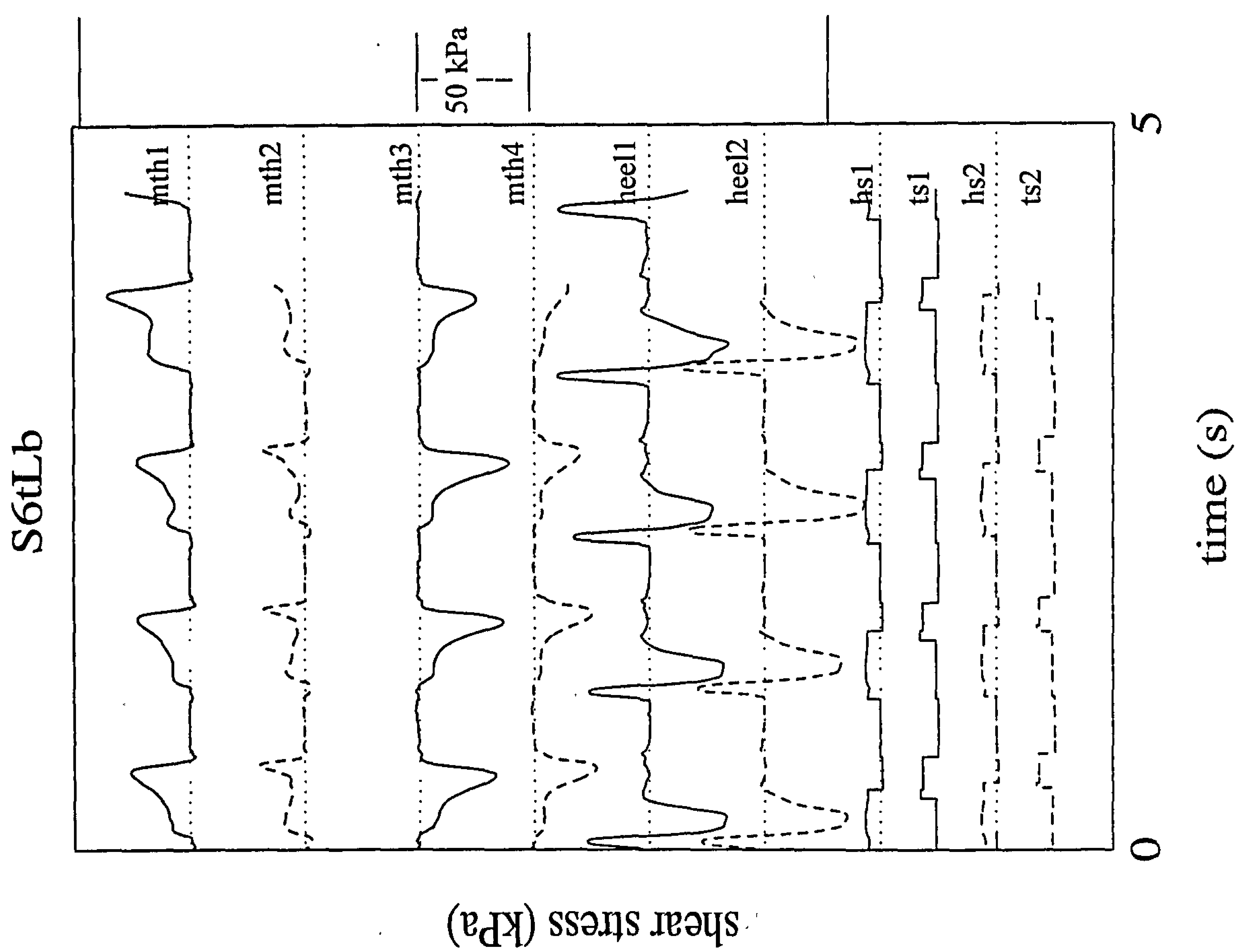

S5tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

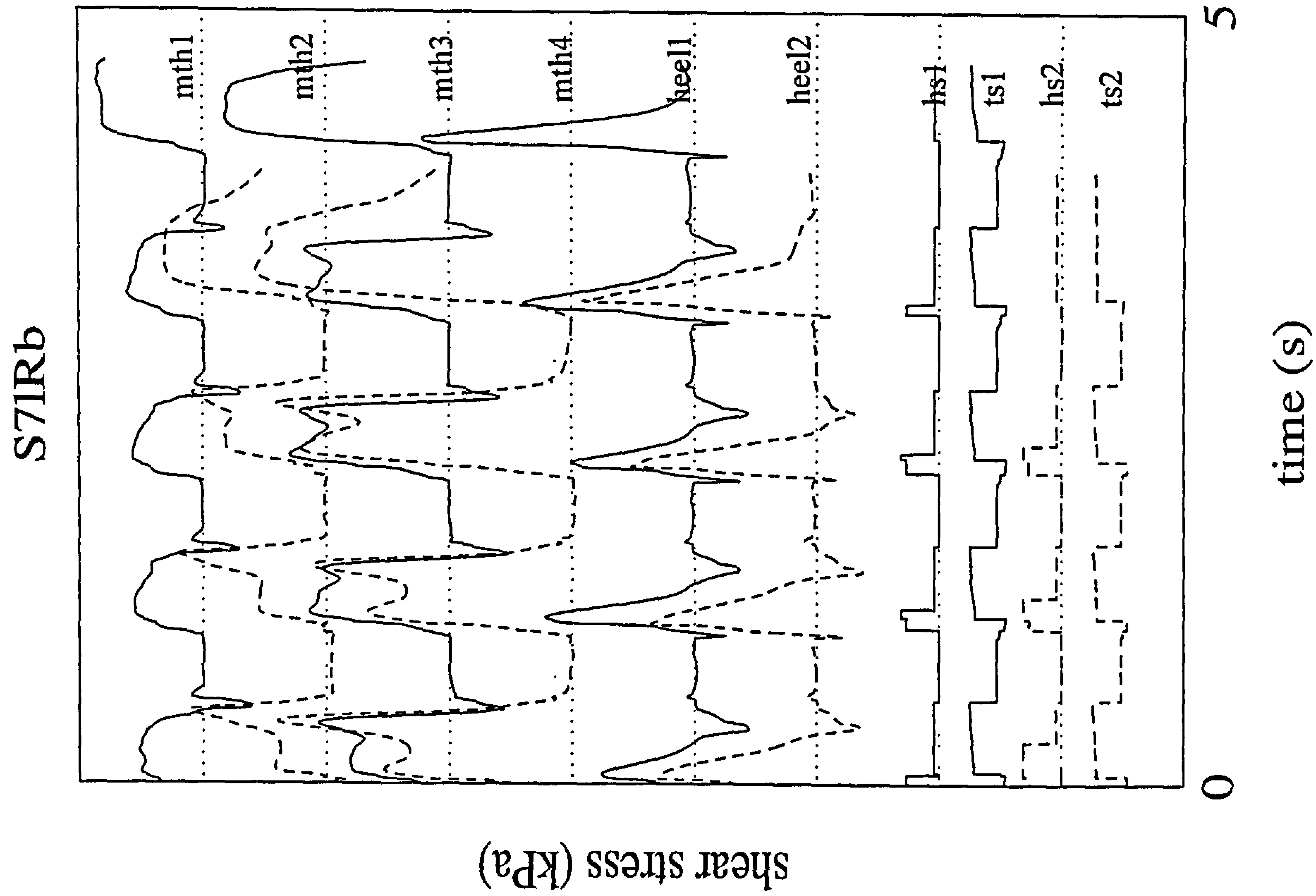

S5tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

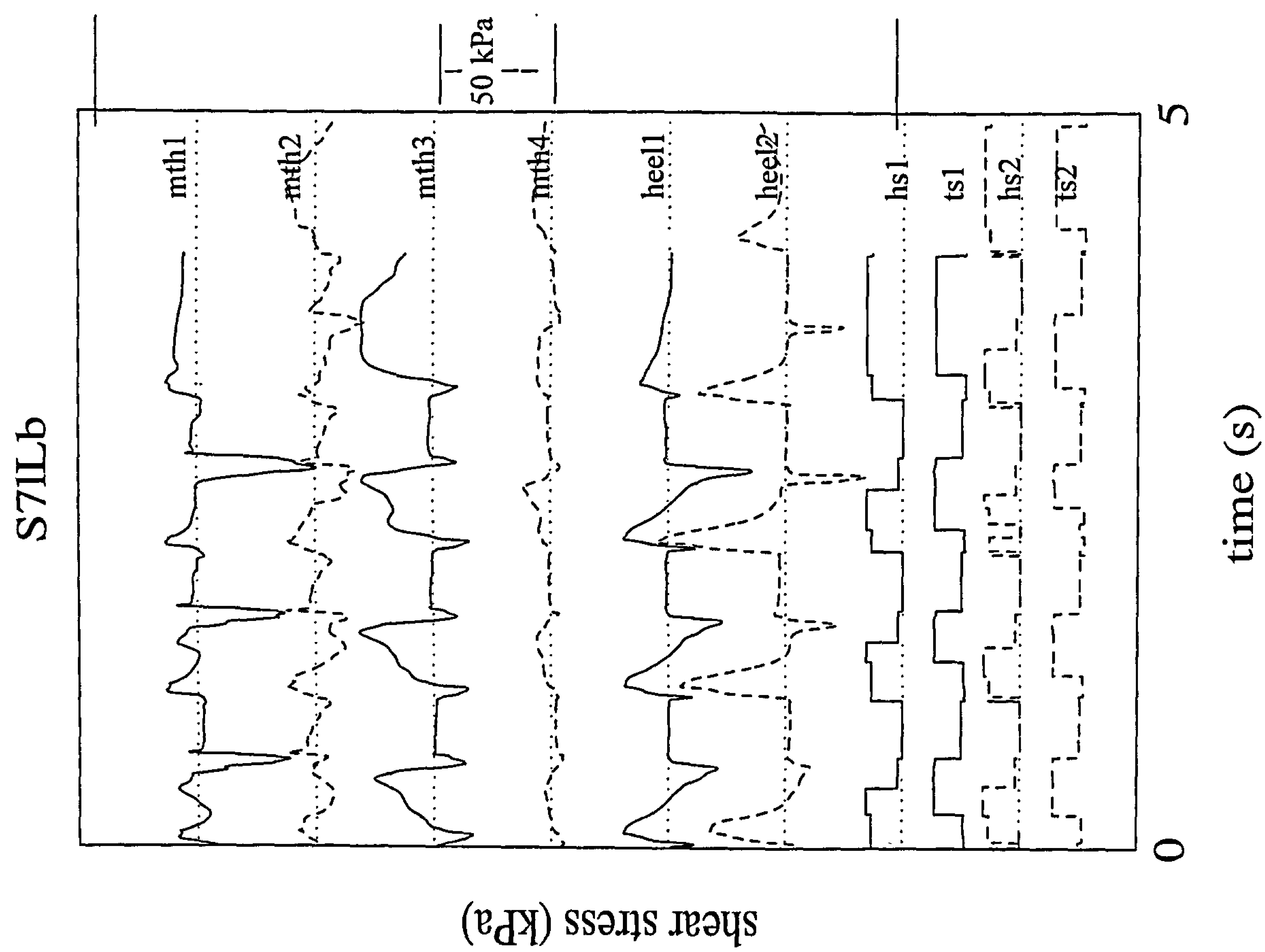

S61Rb
shear stress (kPa)
time (s)
0
5
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2

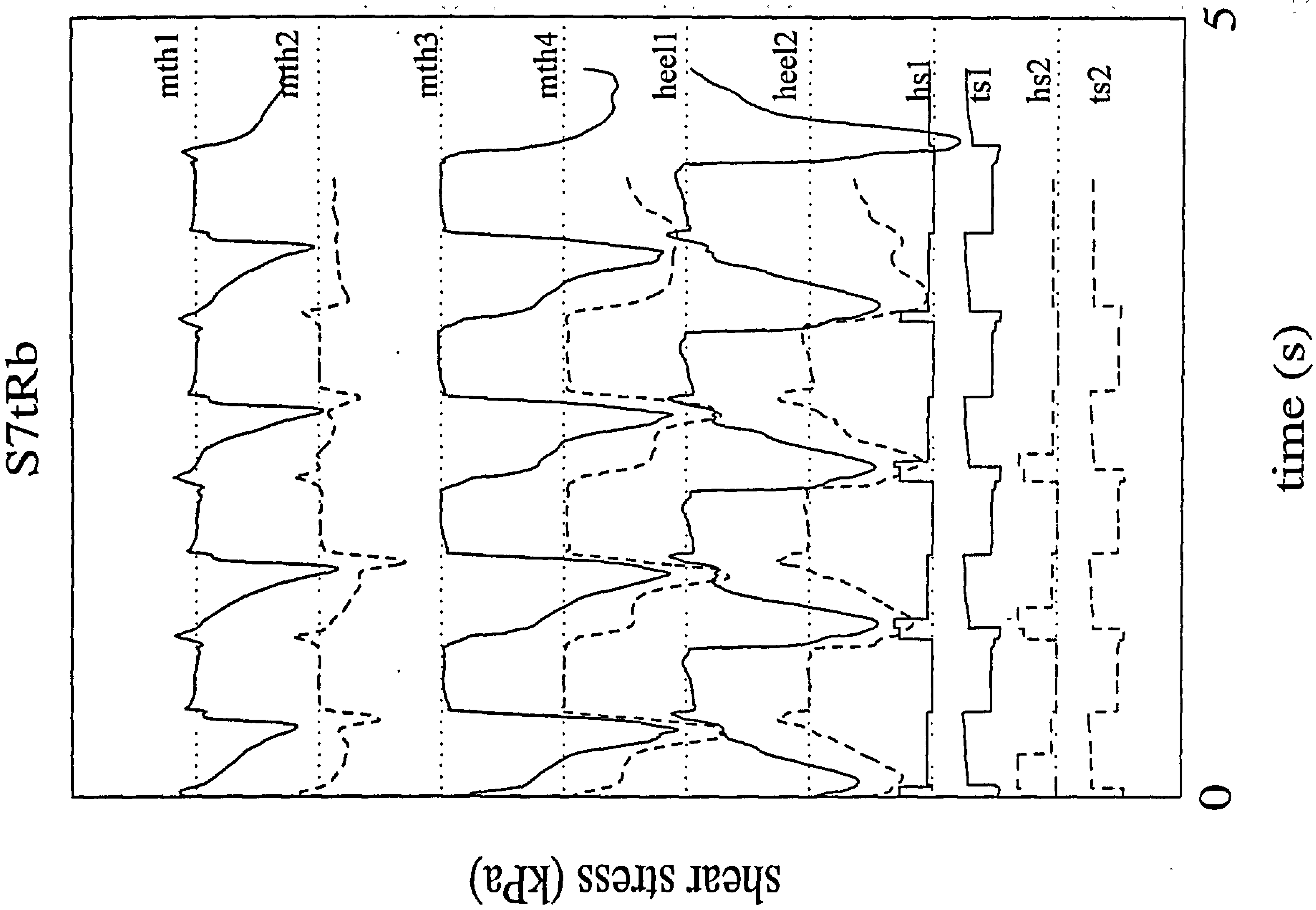

S61Lb
shear stress (kPa)
time (s)
0
5
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2

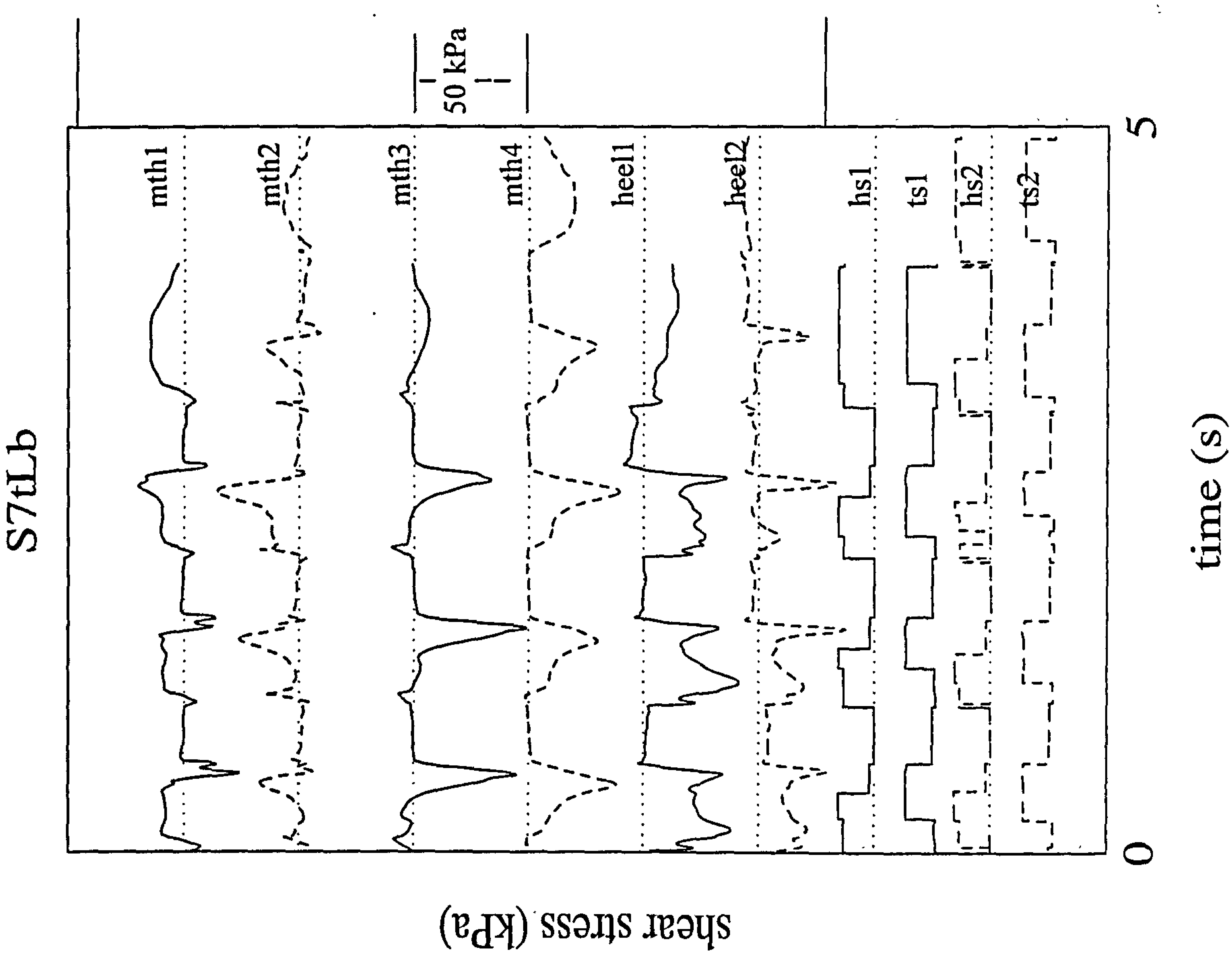

S6tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

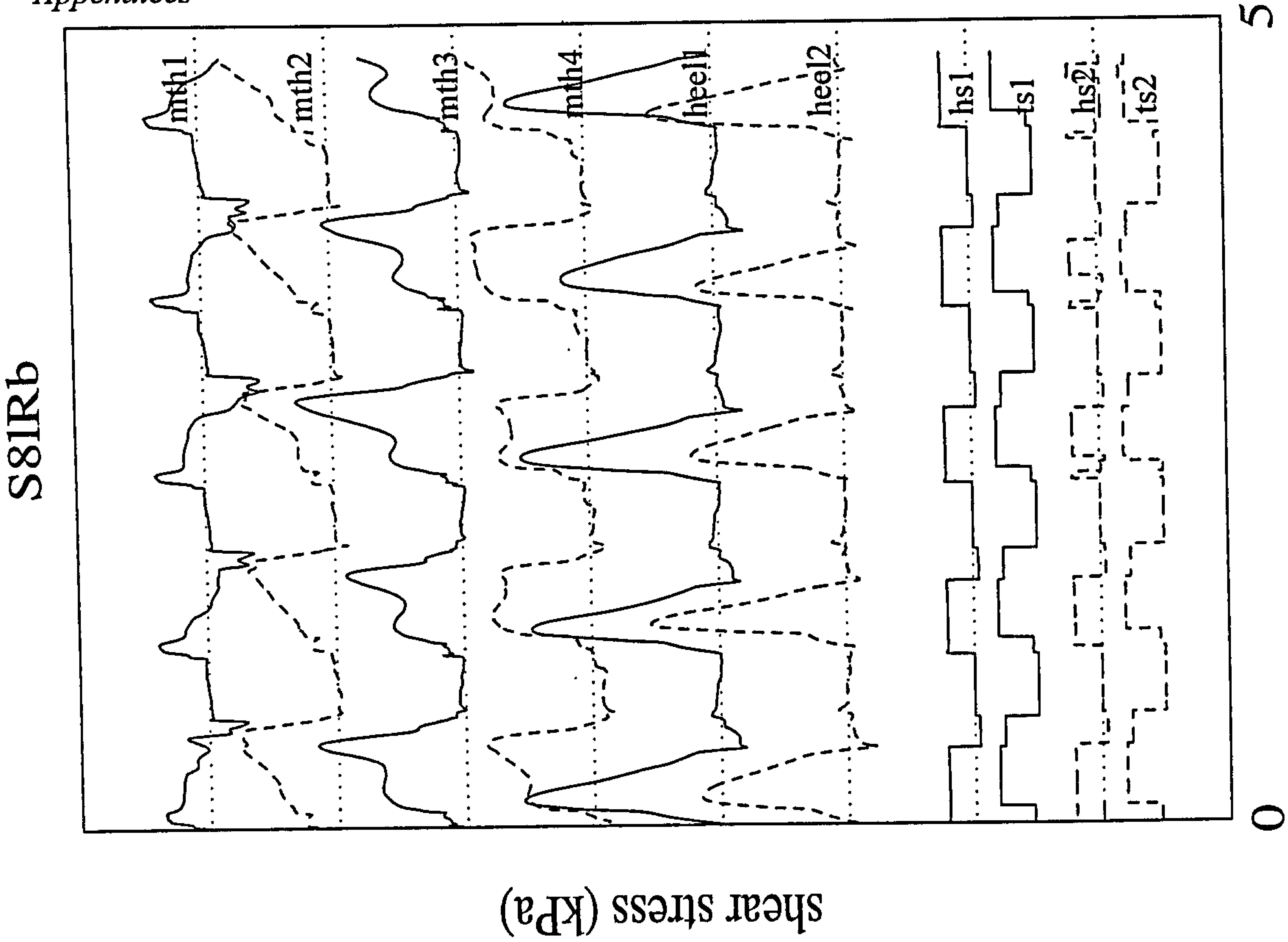

S6tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

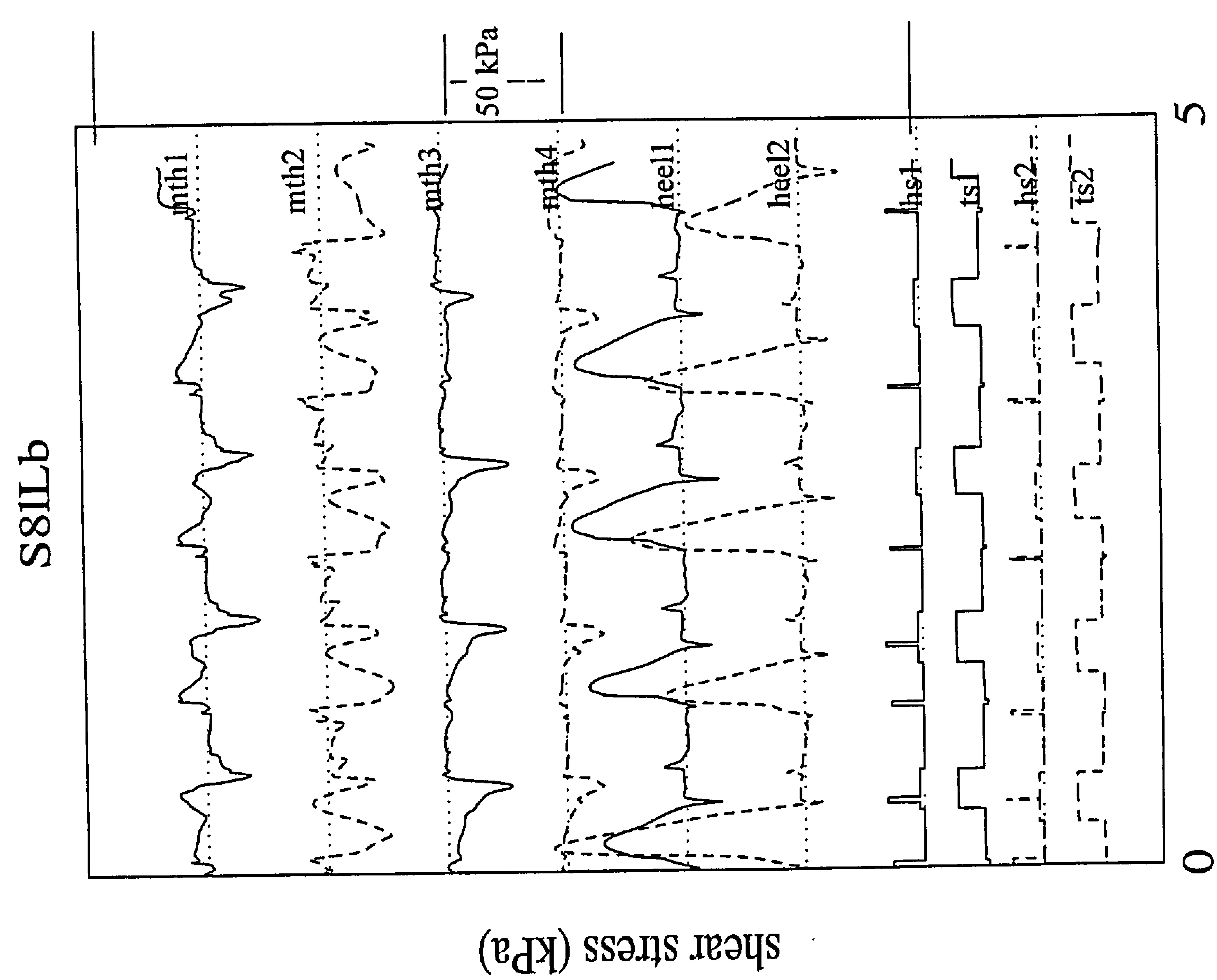

S71Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

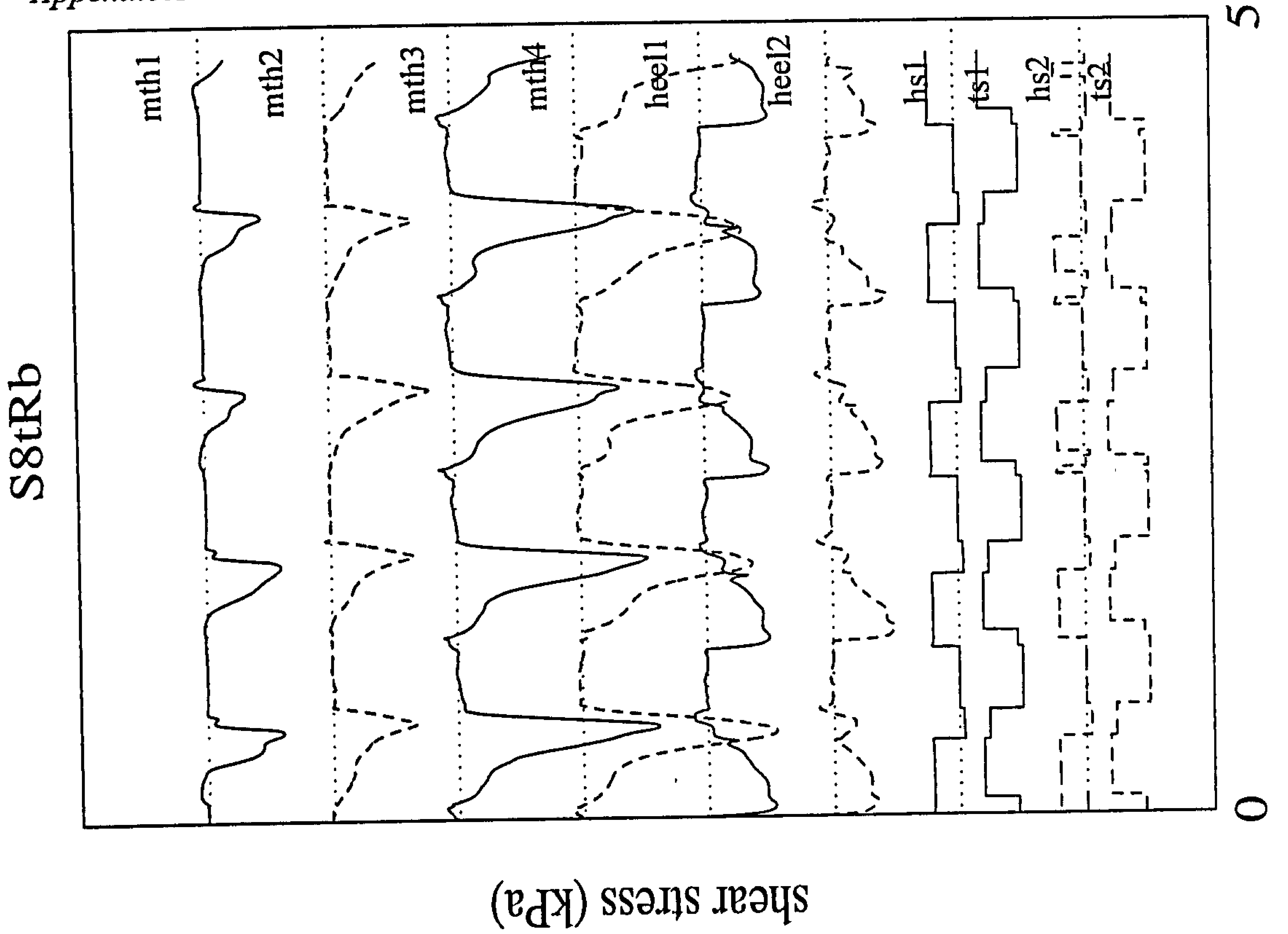

S71Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

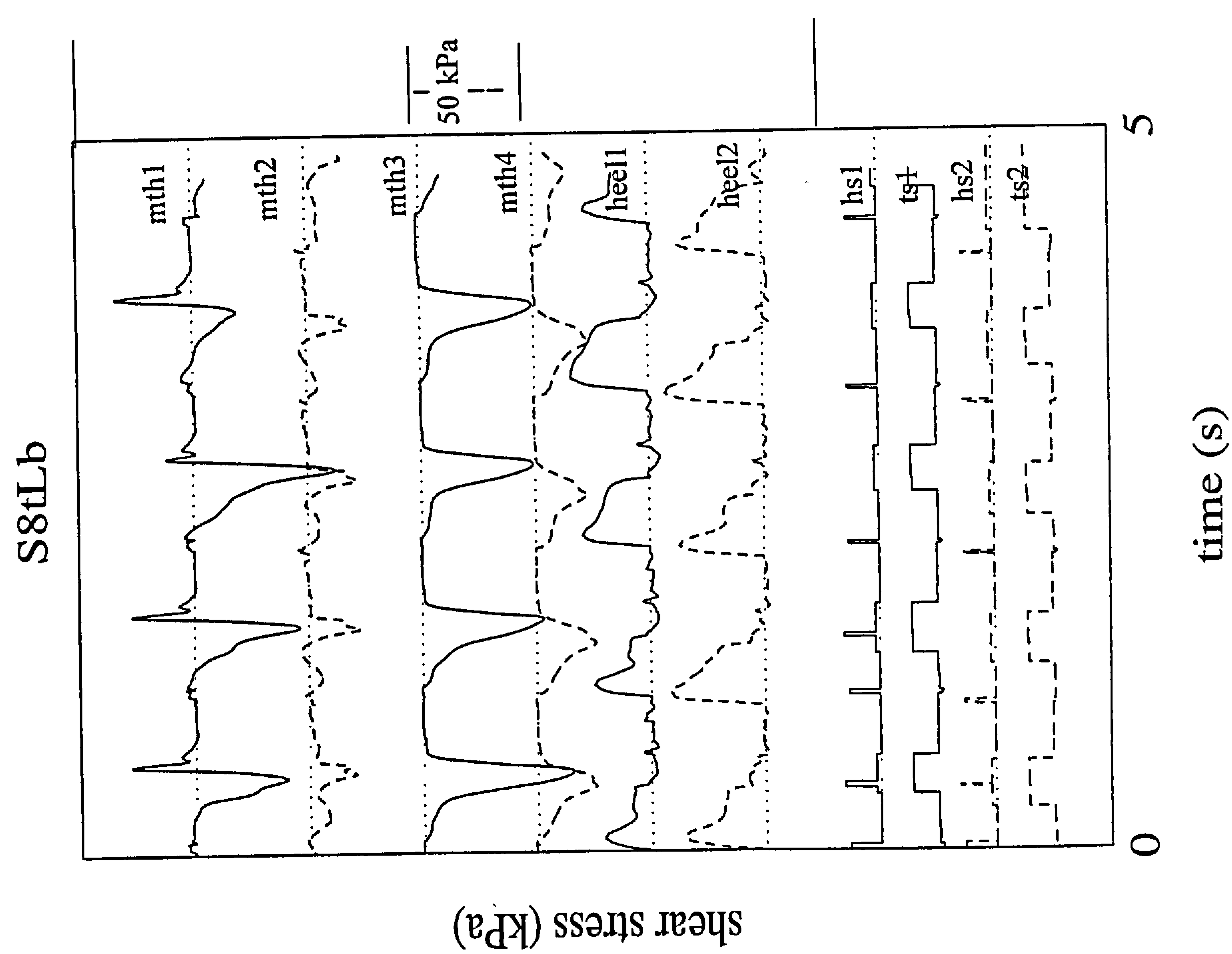

S7tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

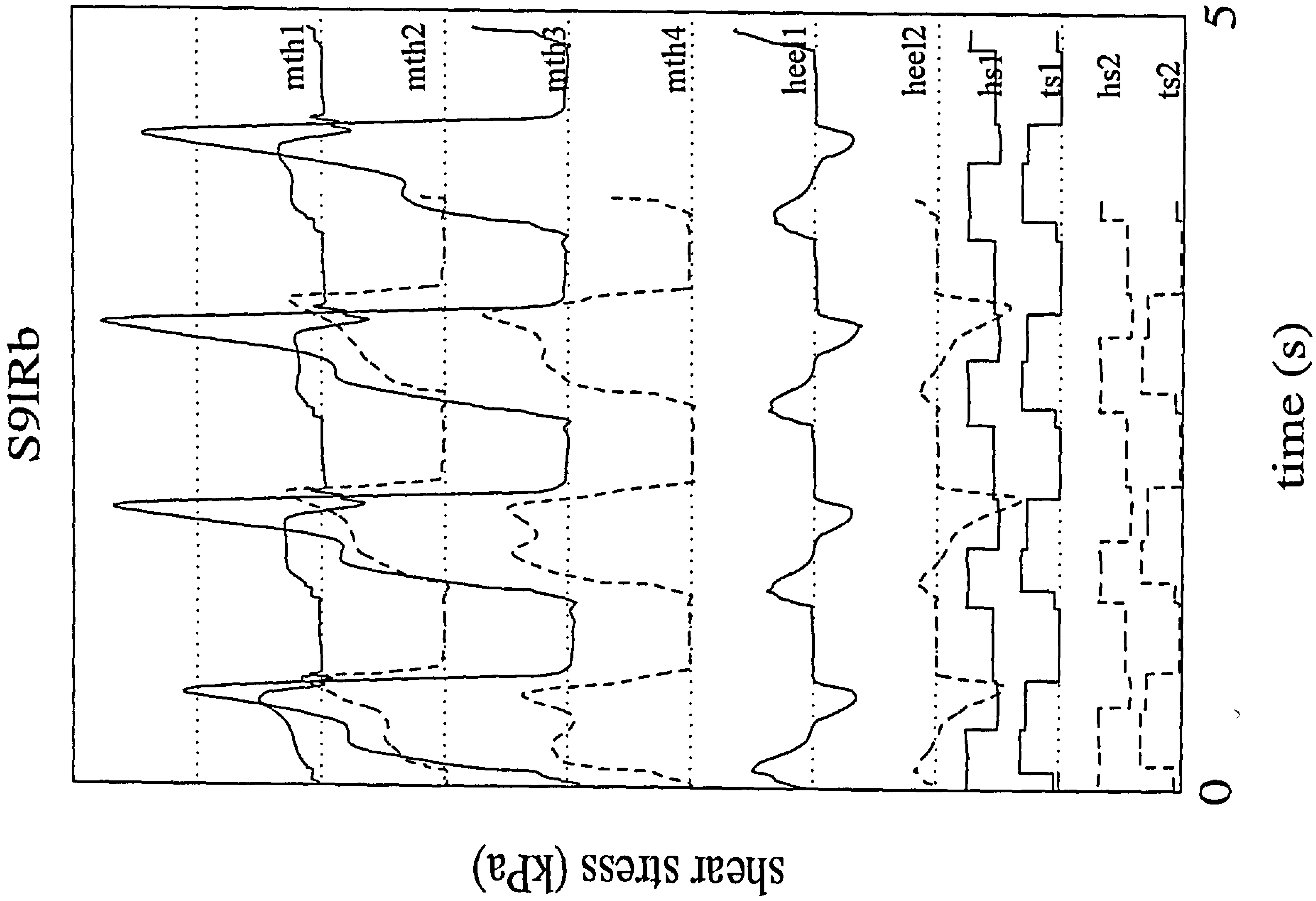

S7tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

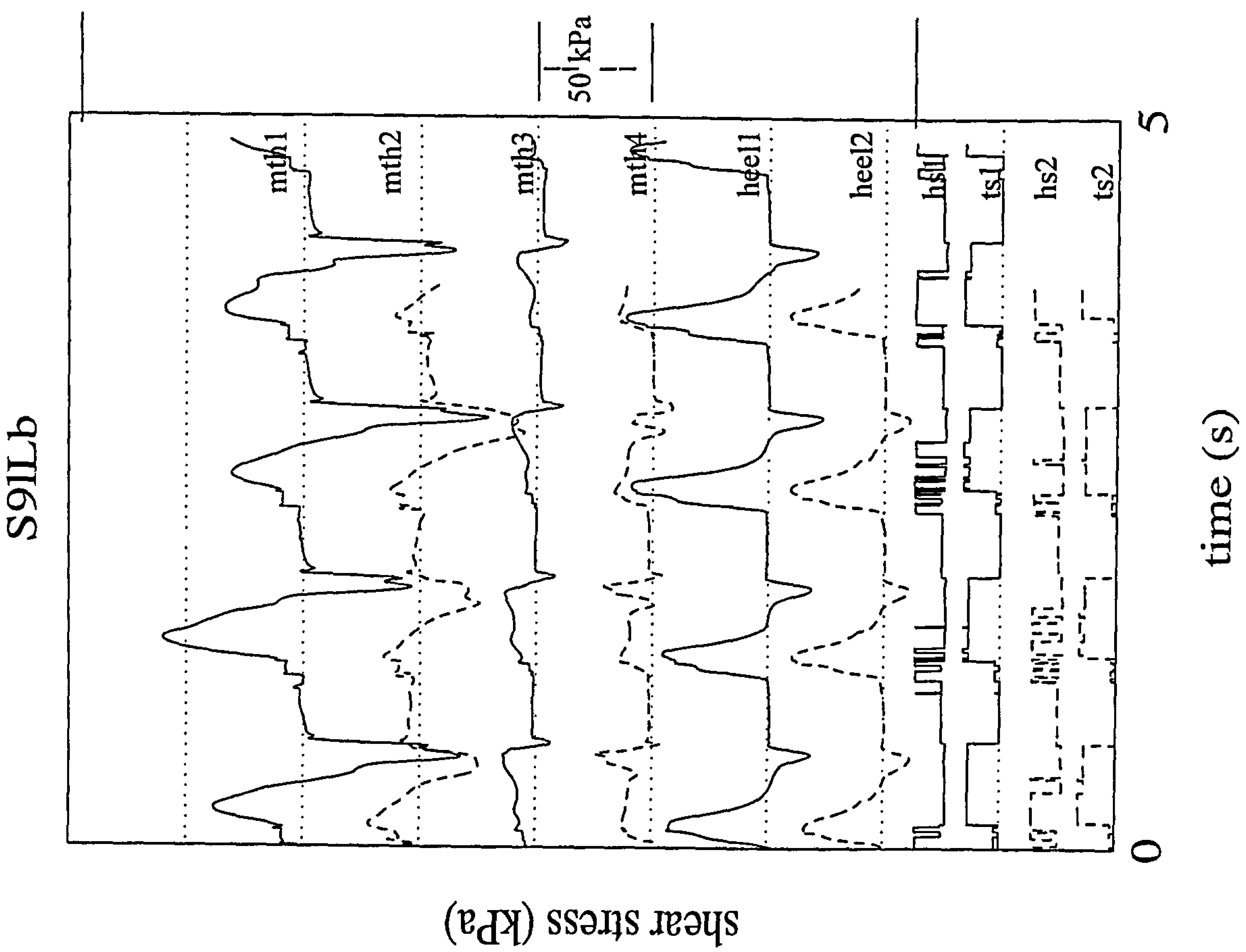

S81Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

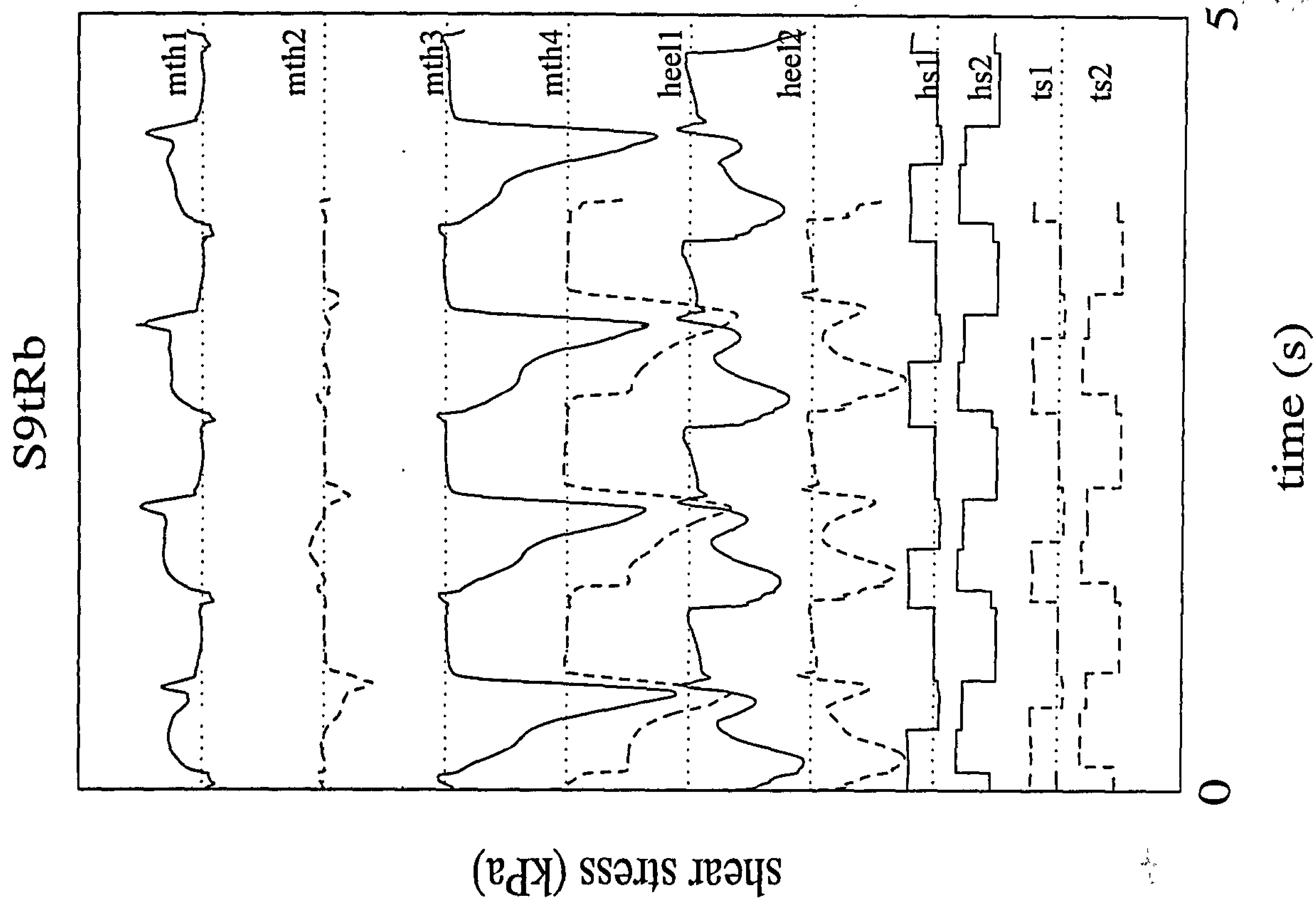

S81Lb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
time (s)
0
5

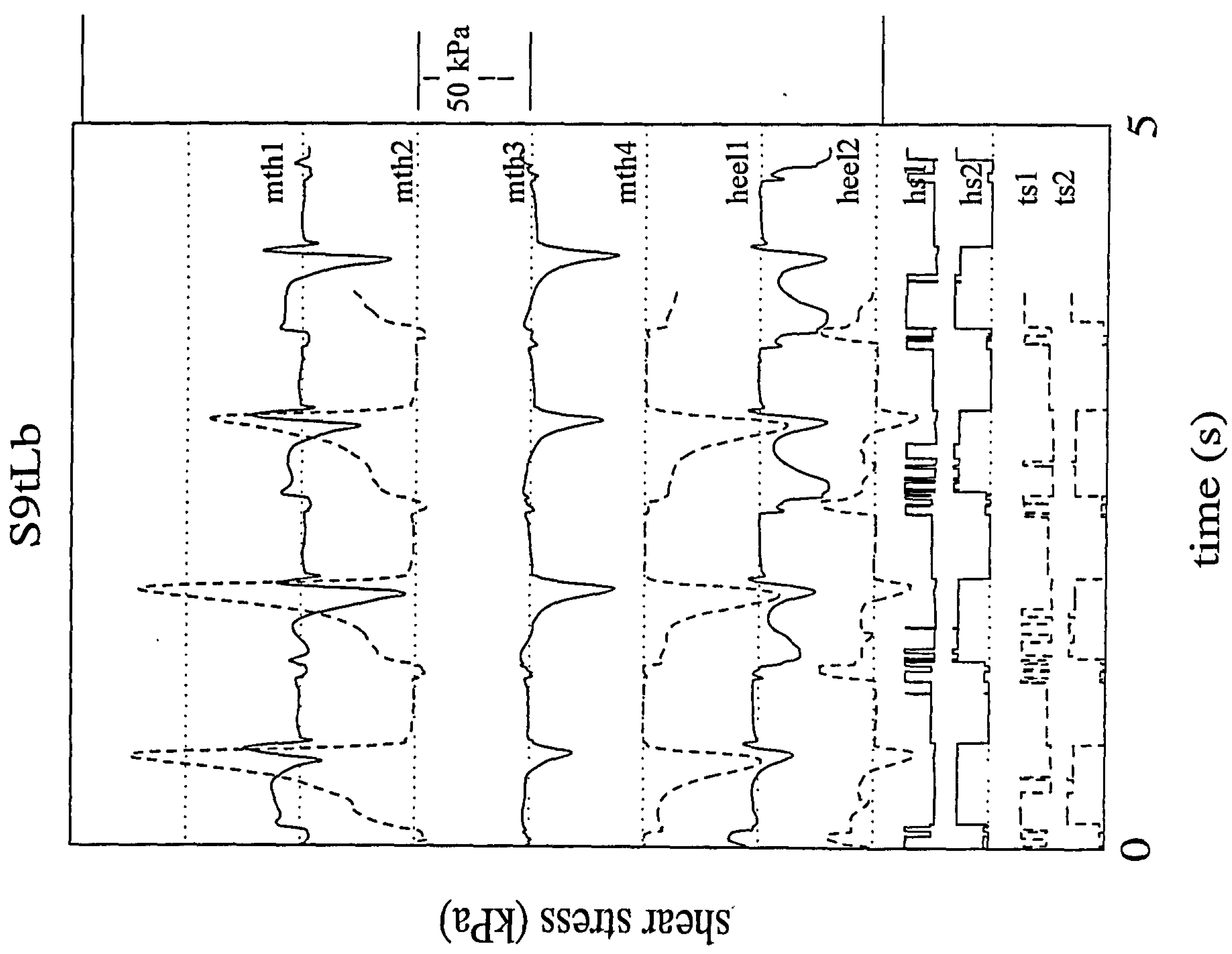

S8tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

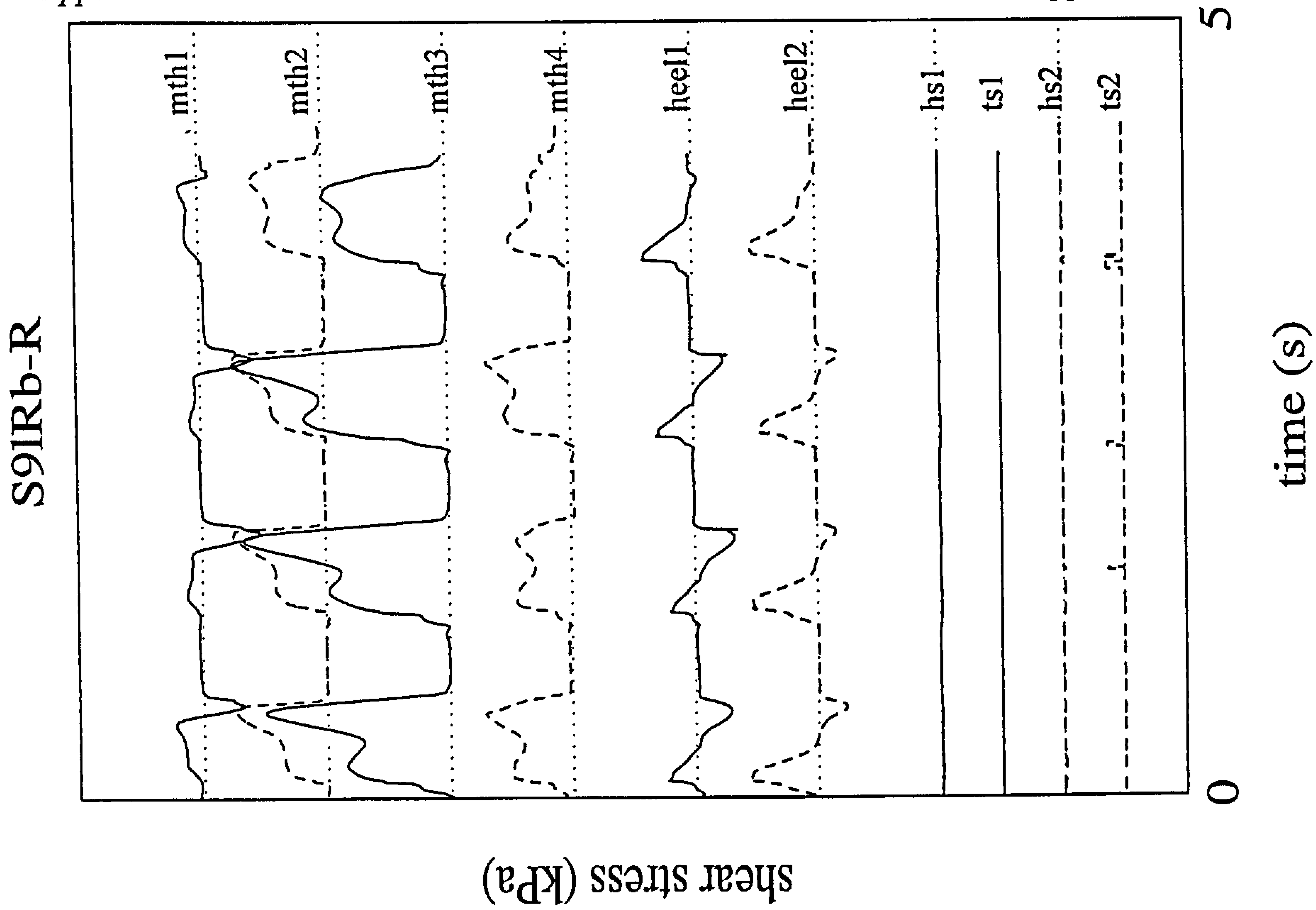

S8tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

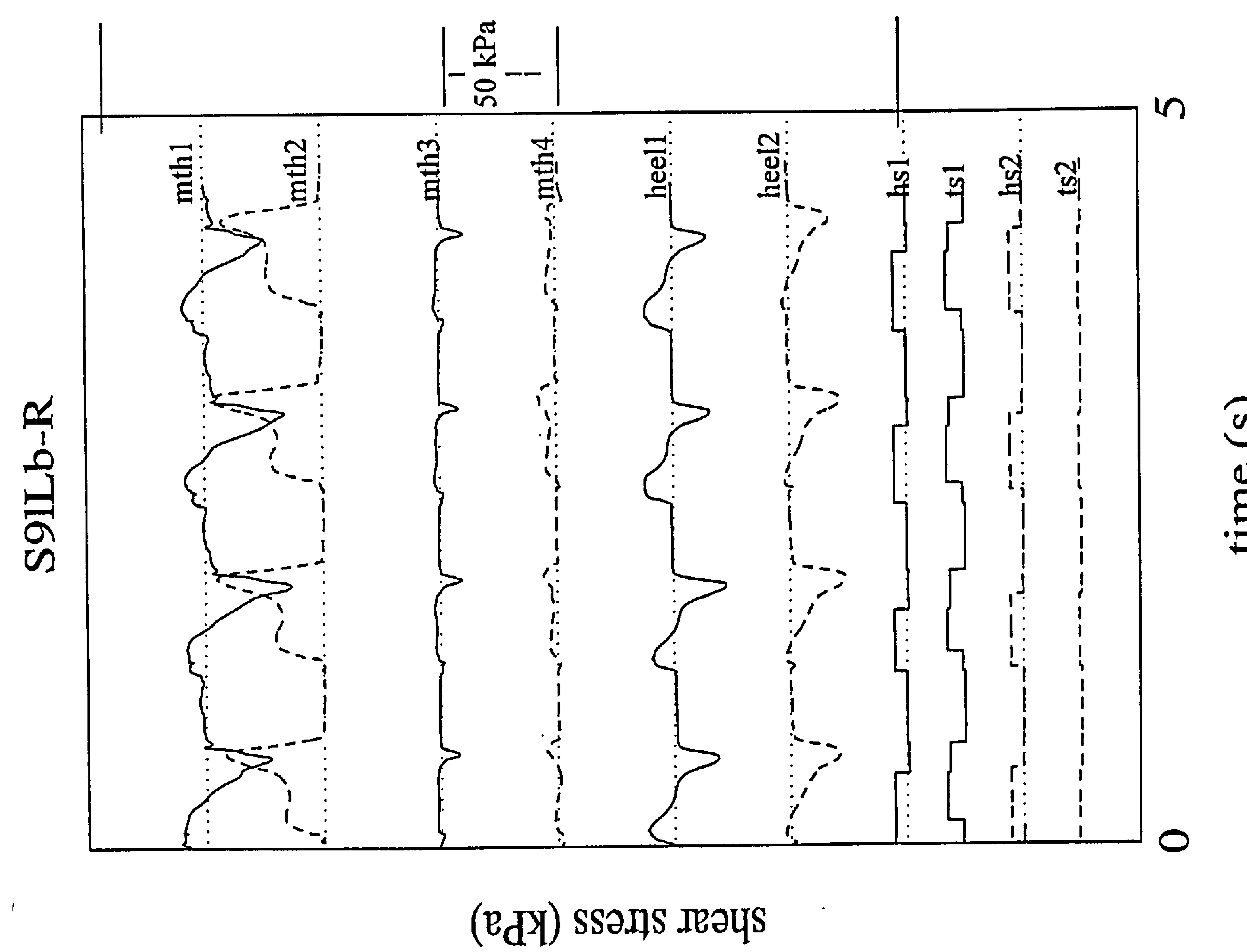

S91Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

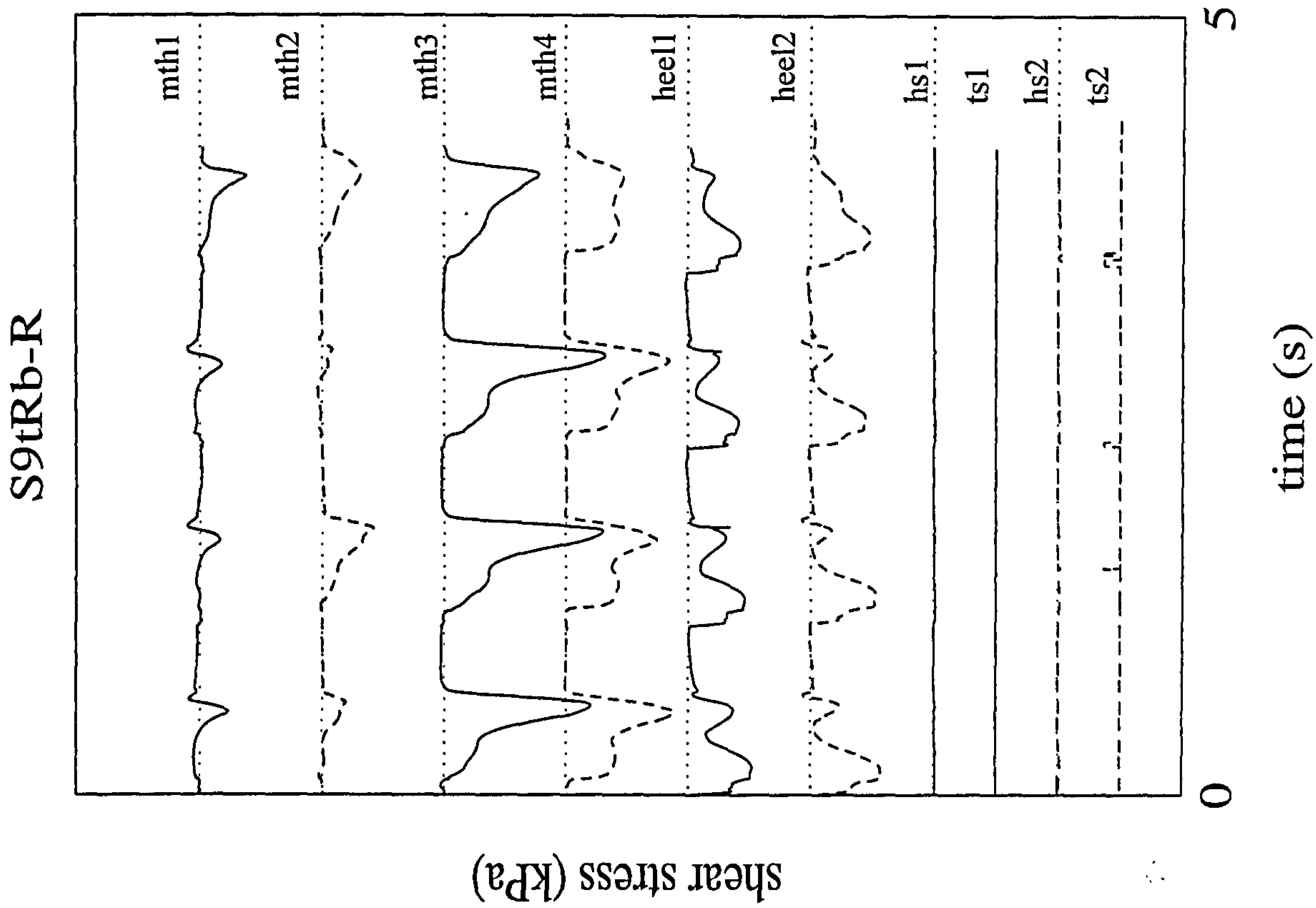

S91Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

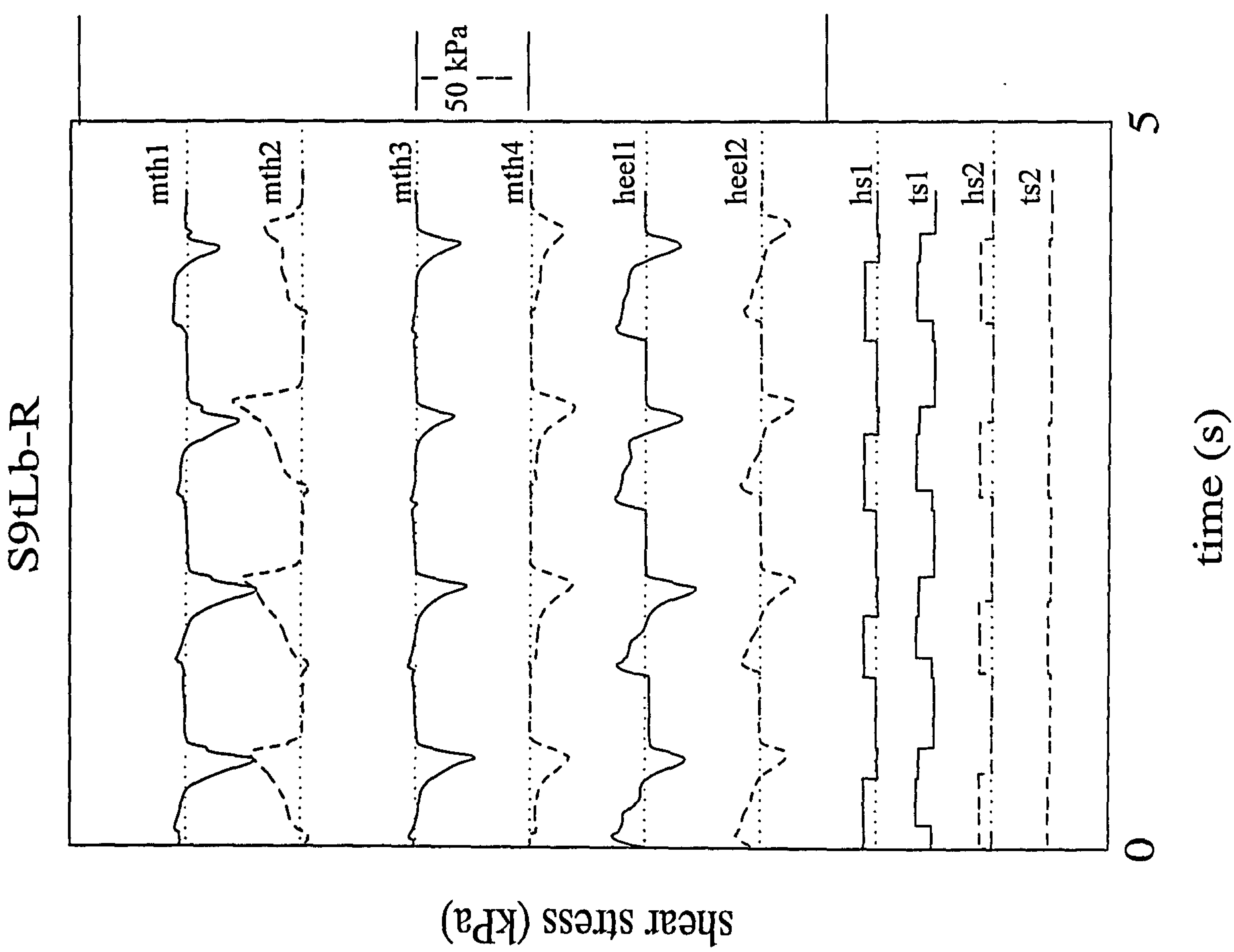

S9tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
hs2
ts1
ts2
0
5
time (s)
shear stress (kPa)

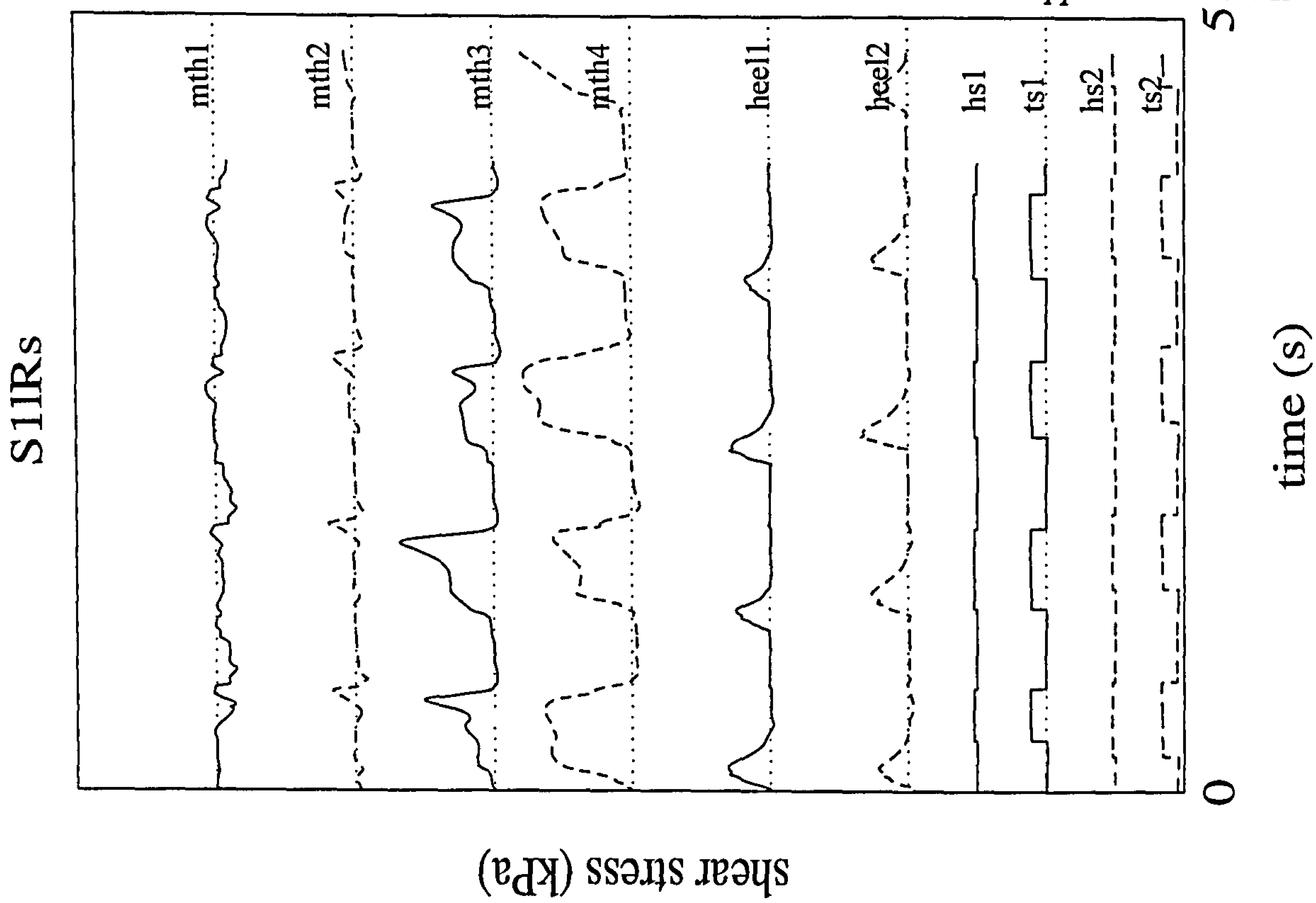

S9tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
hs2
ts1
ts2
0
5
time (s)
shear stress (kPa)

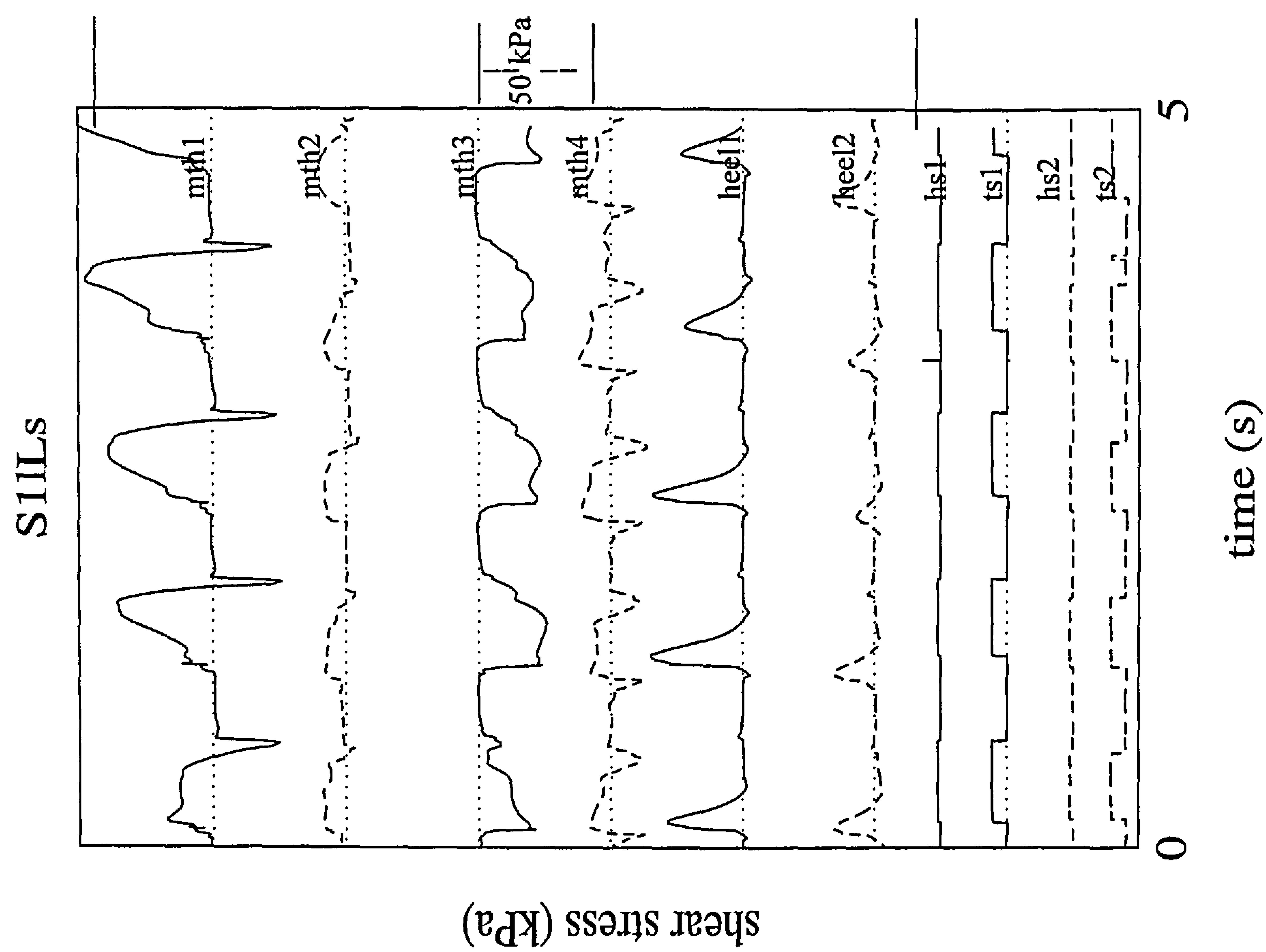

S91Rb-R
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

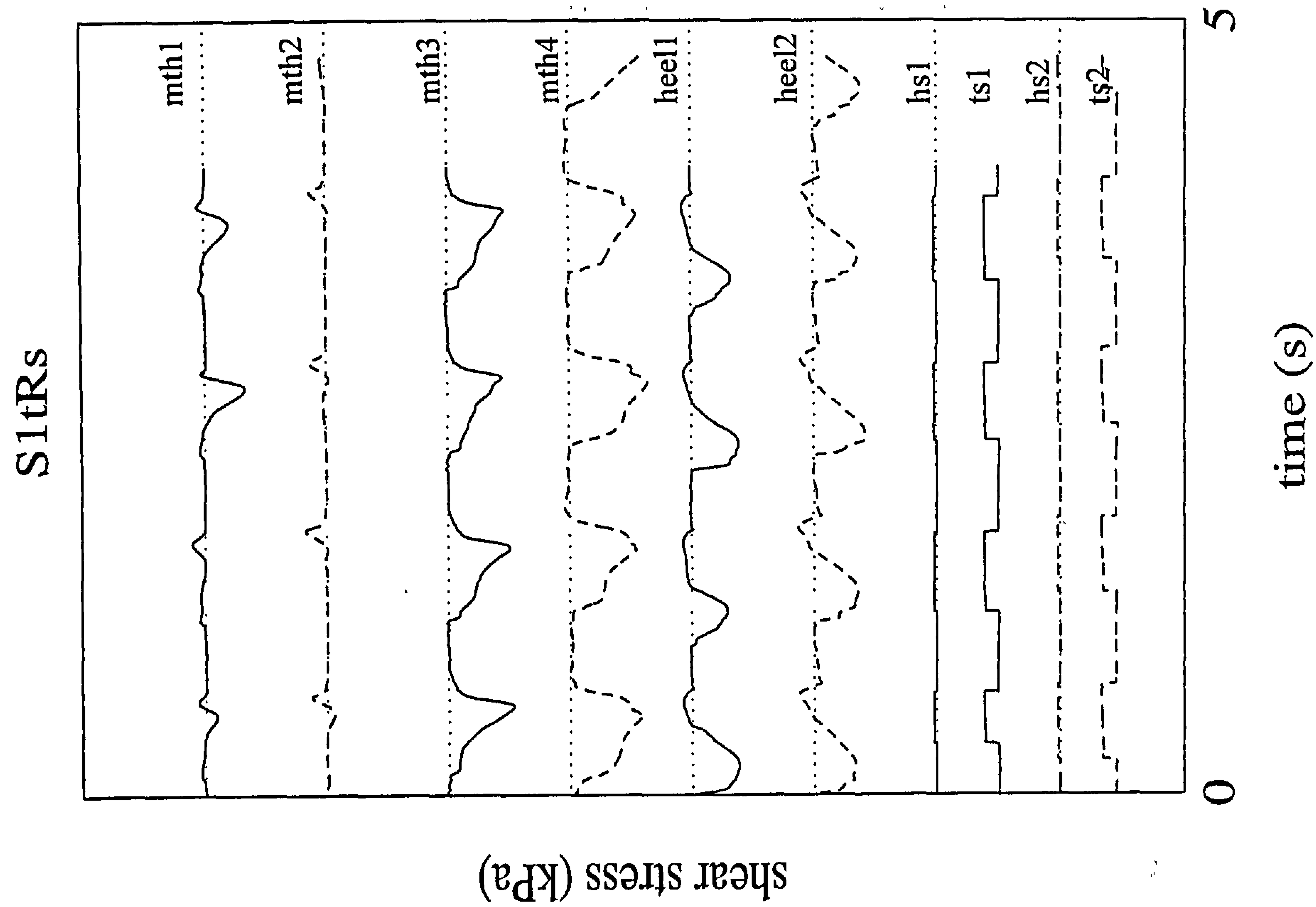

S91Lb-R
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

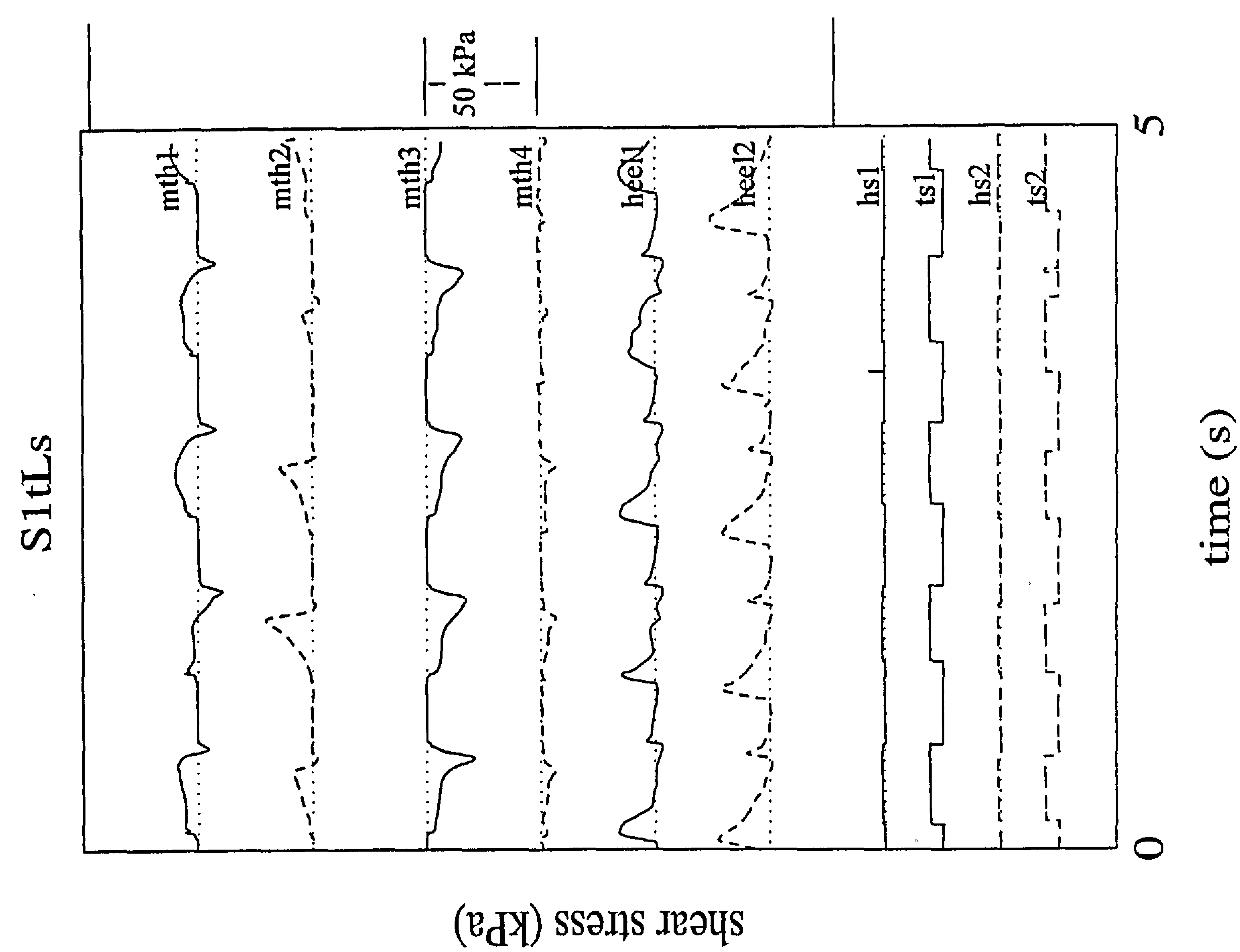

S9tRb-R
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

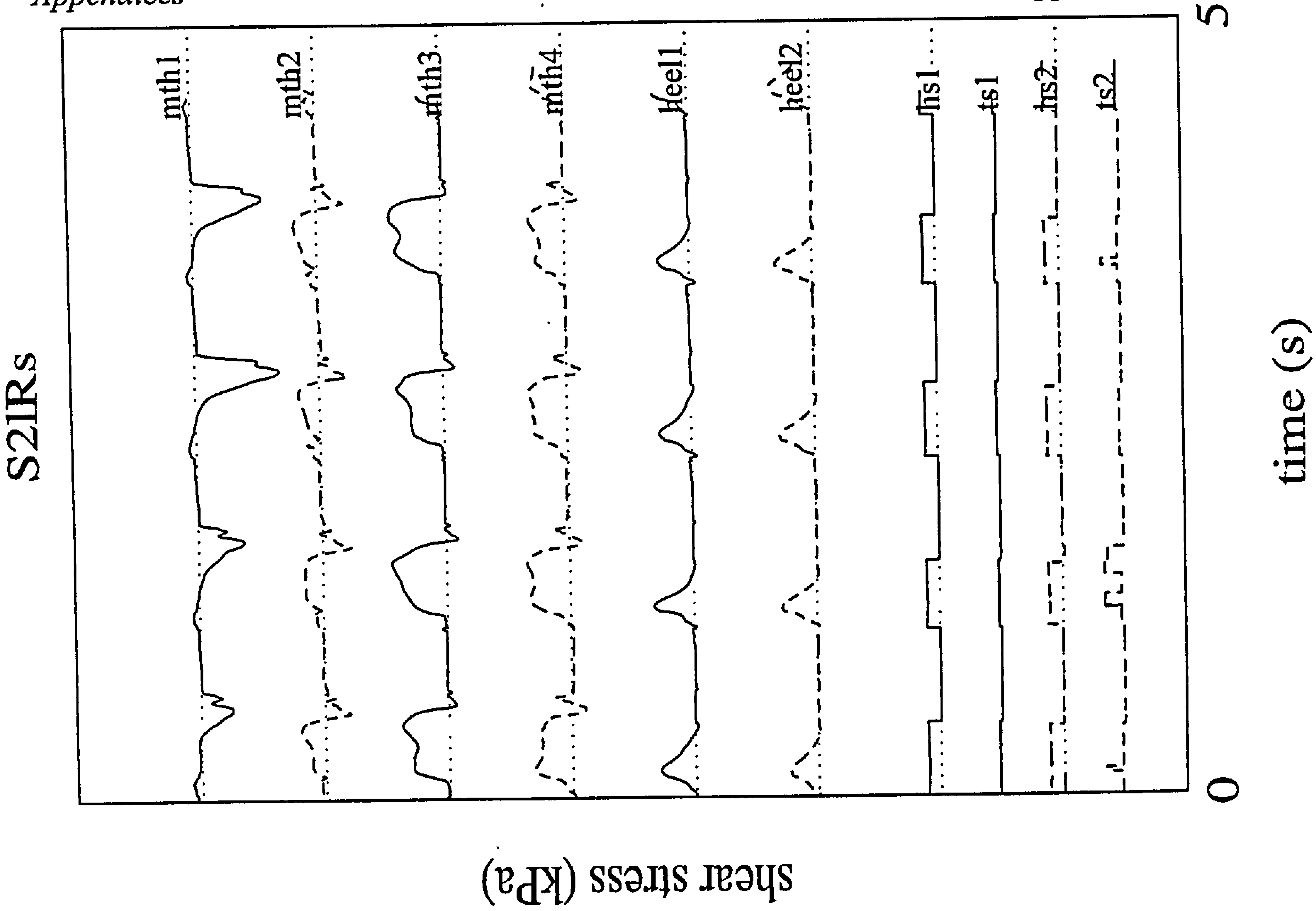

S9tLb-R
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
time (s)
0
5

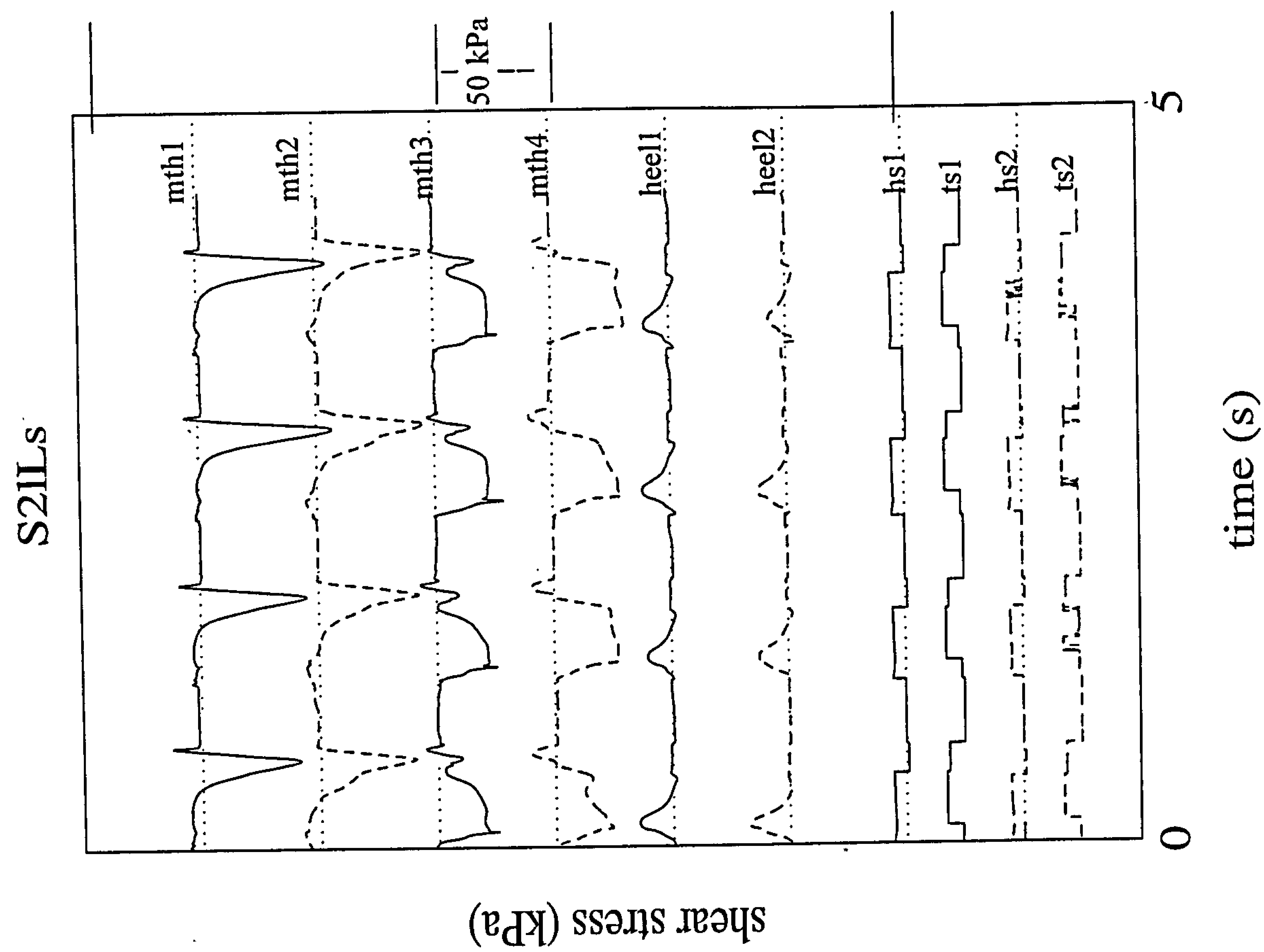

S11Rs
shear stress (kPa)
time (s)
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5

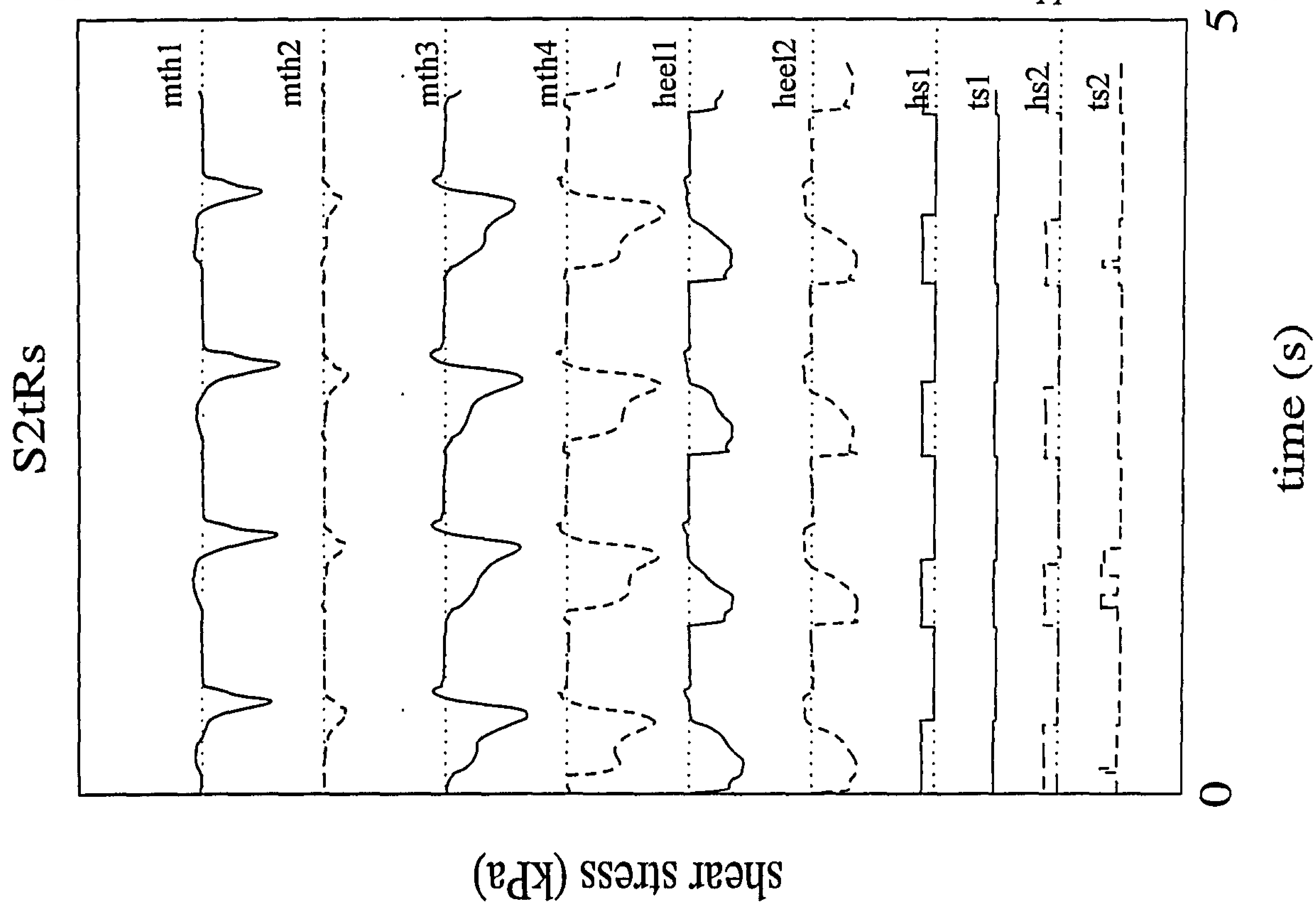

S11Ls
shear stress (kPa)
time (s)
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5

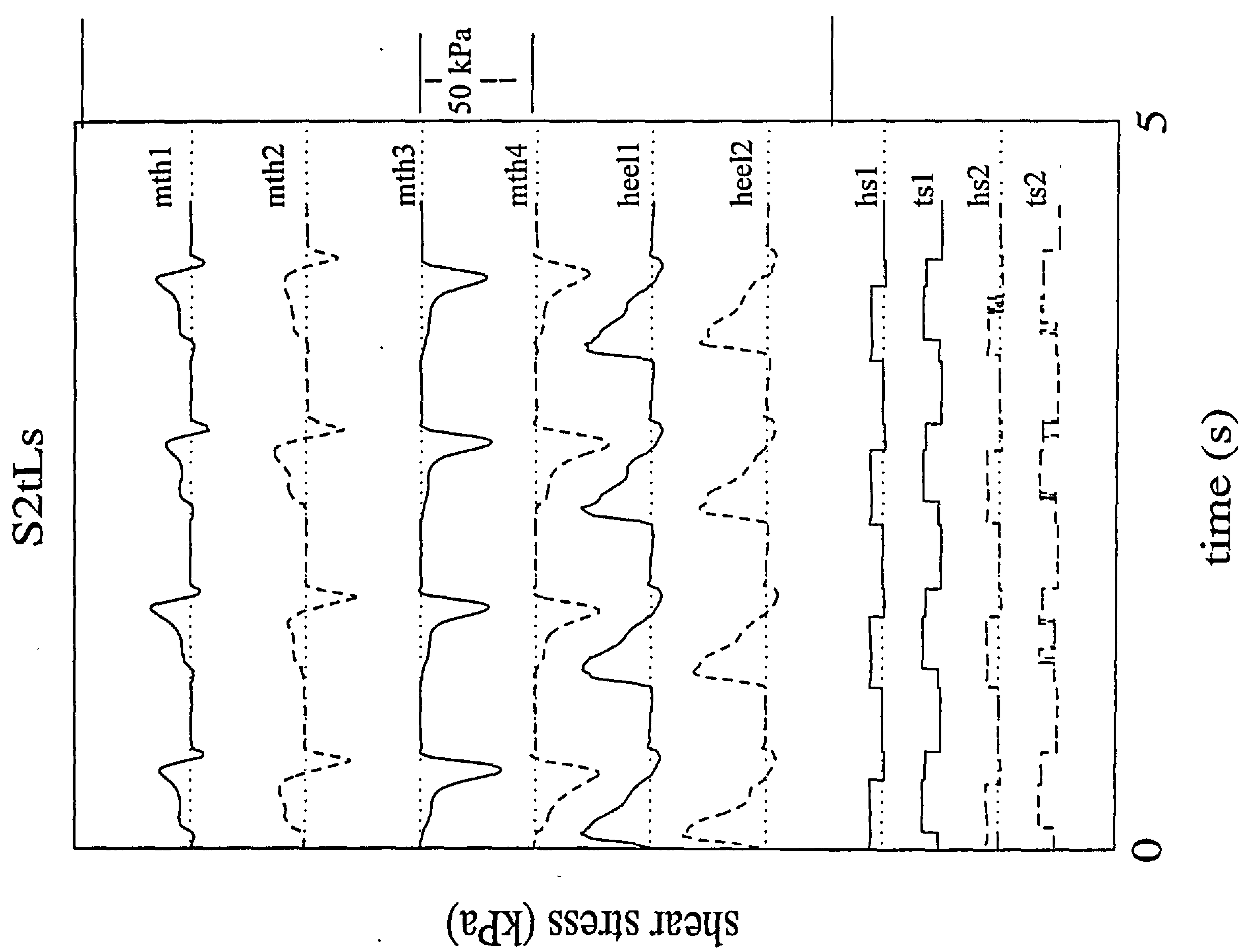

S1tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

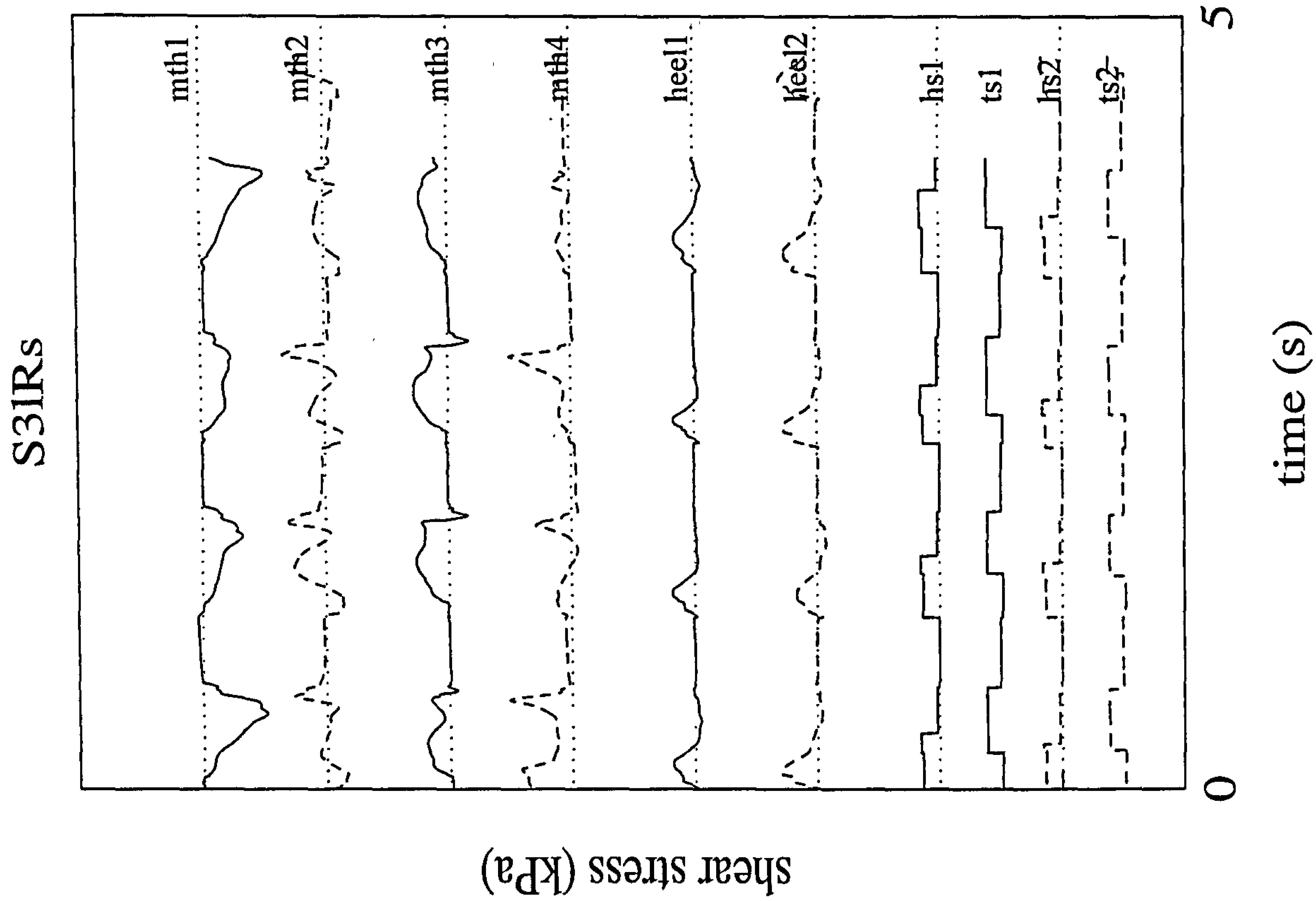

S1tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

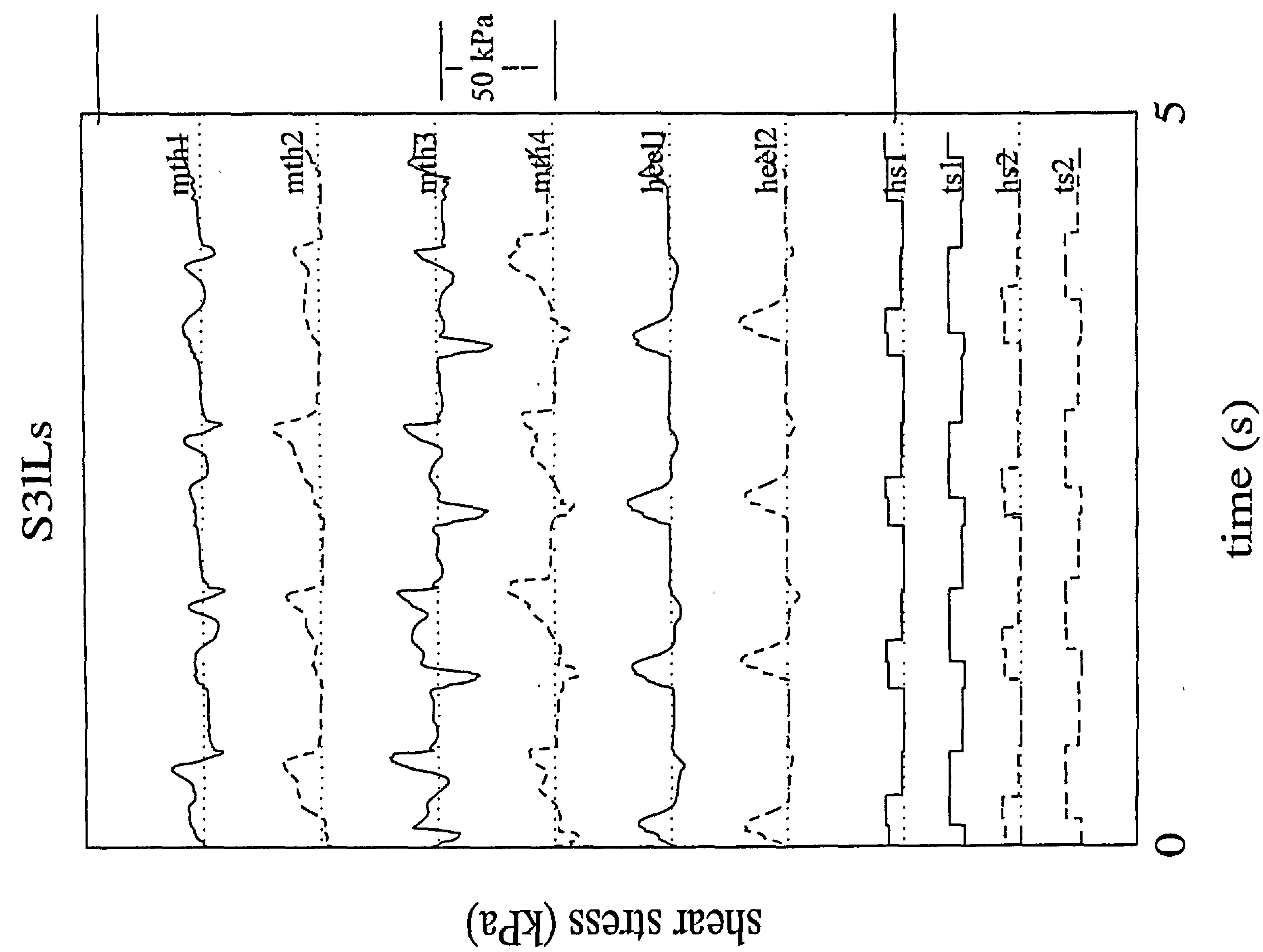

S21Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

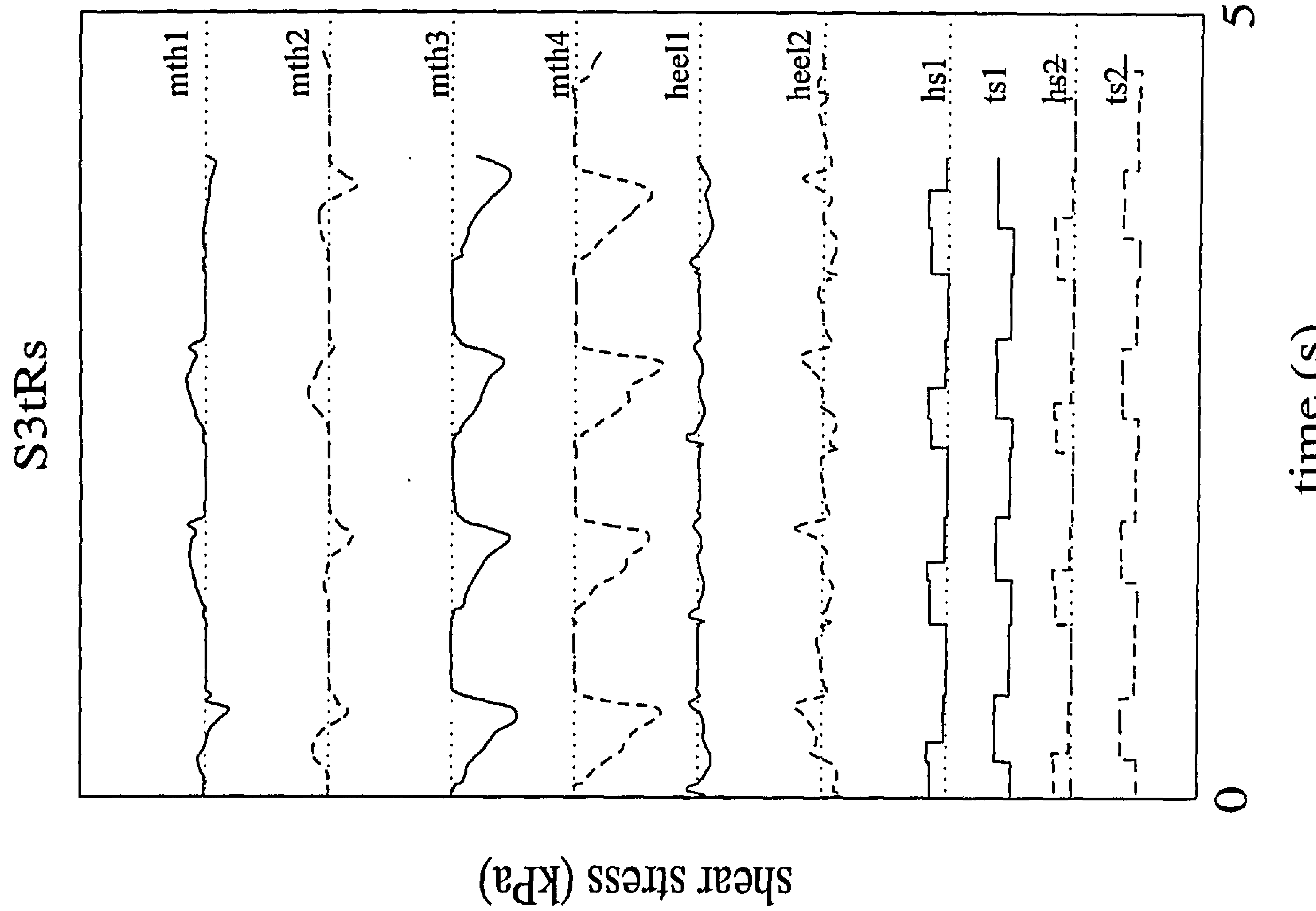

S21Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

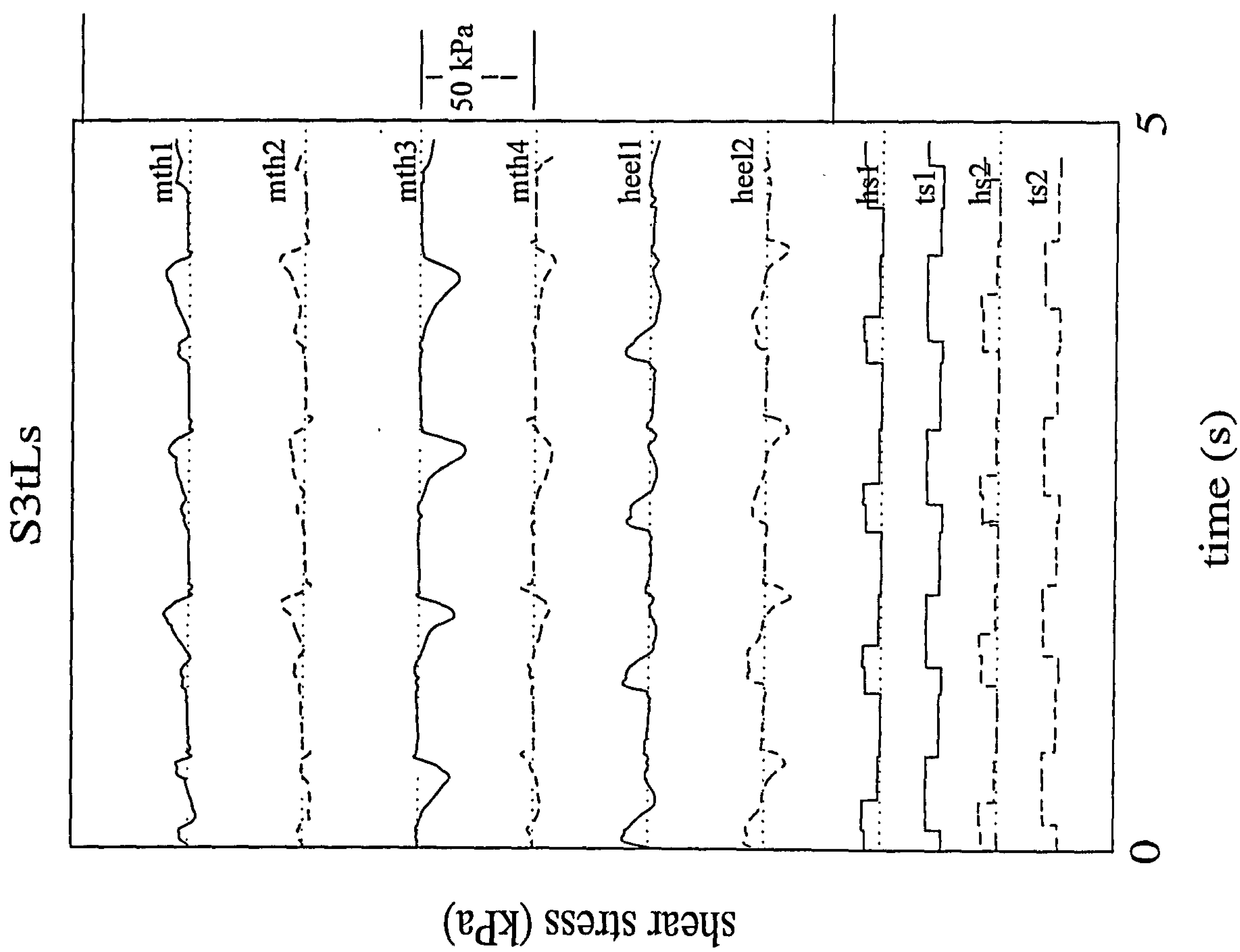

S2tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

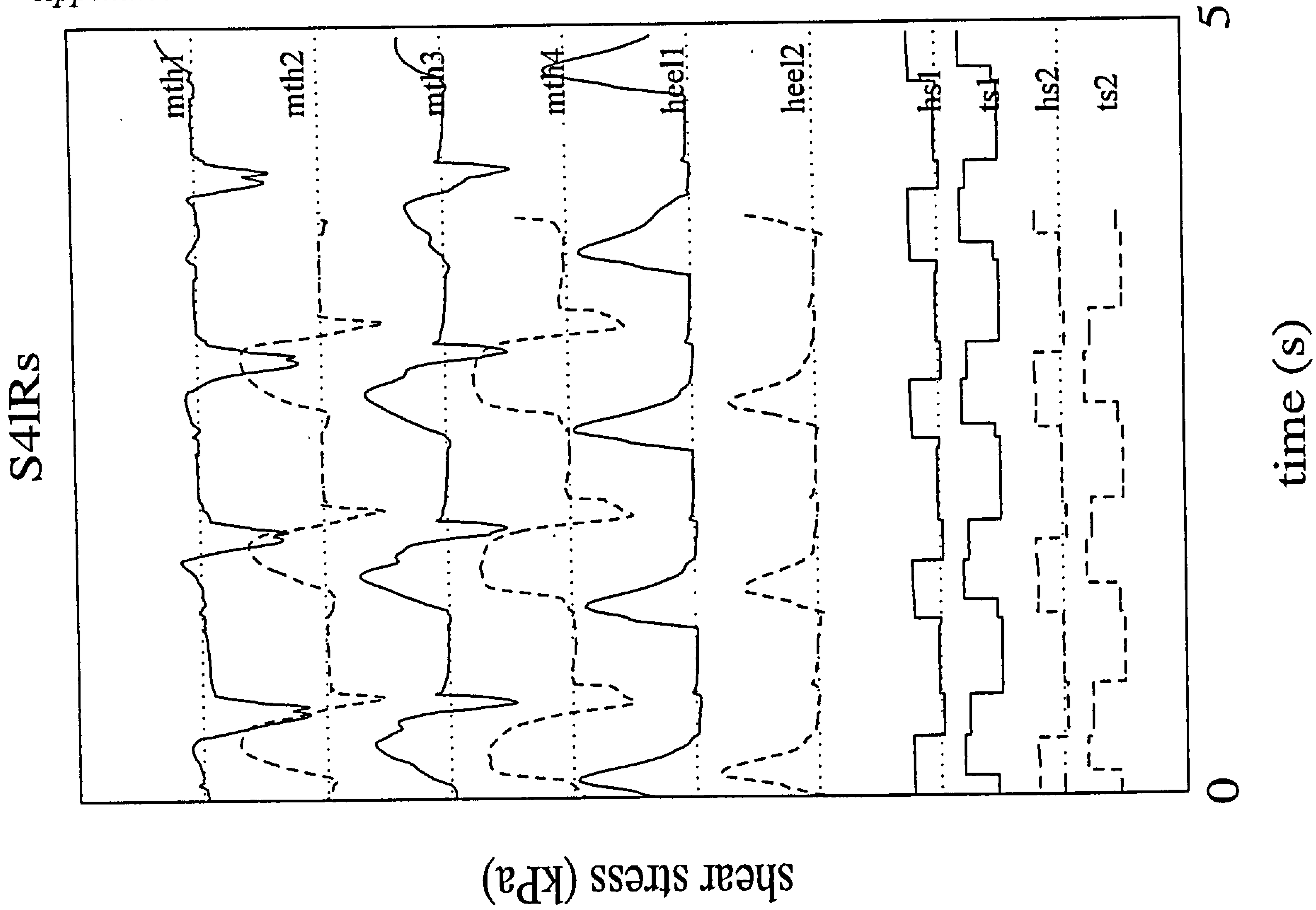

S2tLs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
0
5
time (s)
shear stress (kPa)

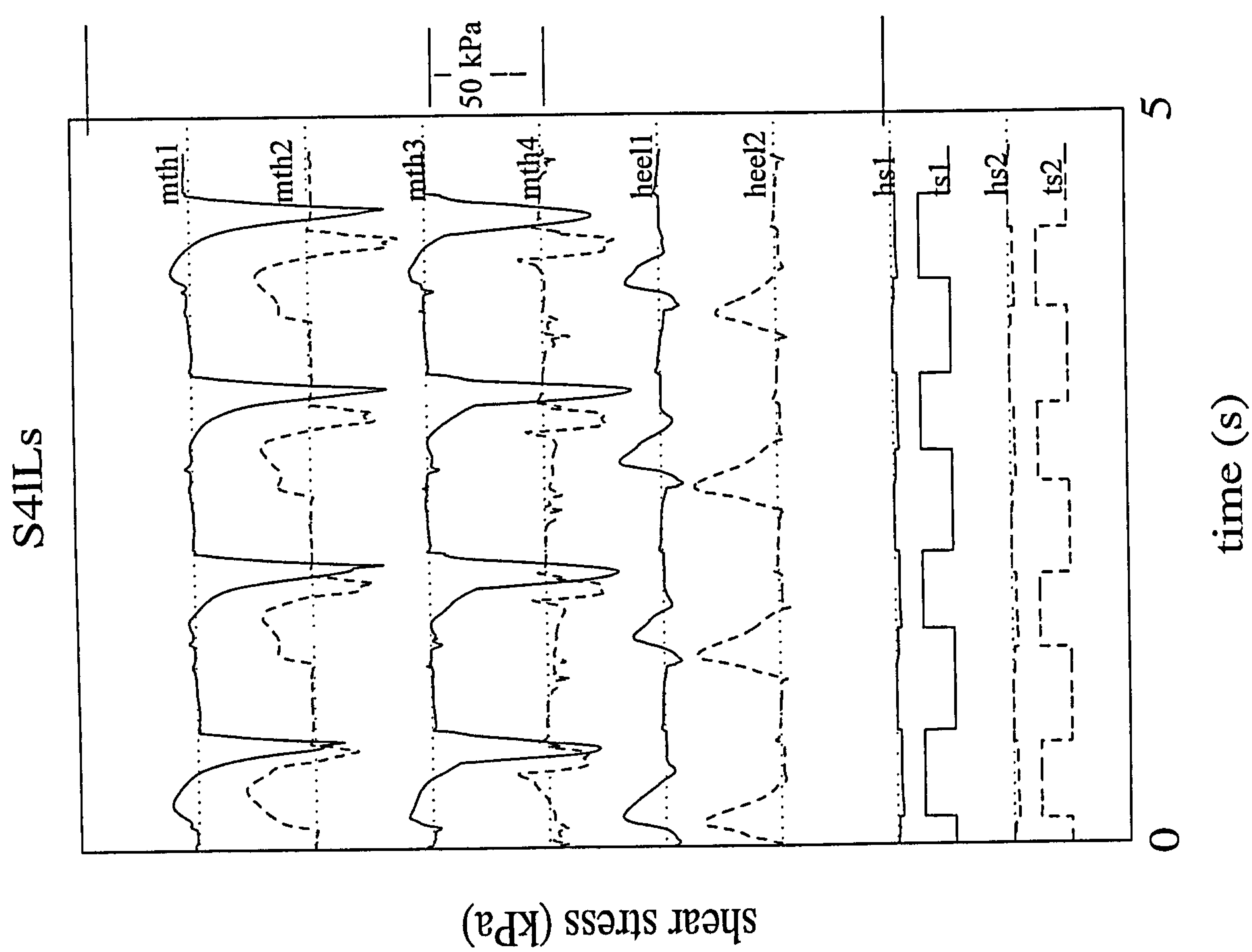

S31Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

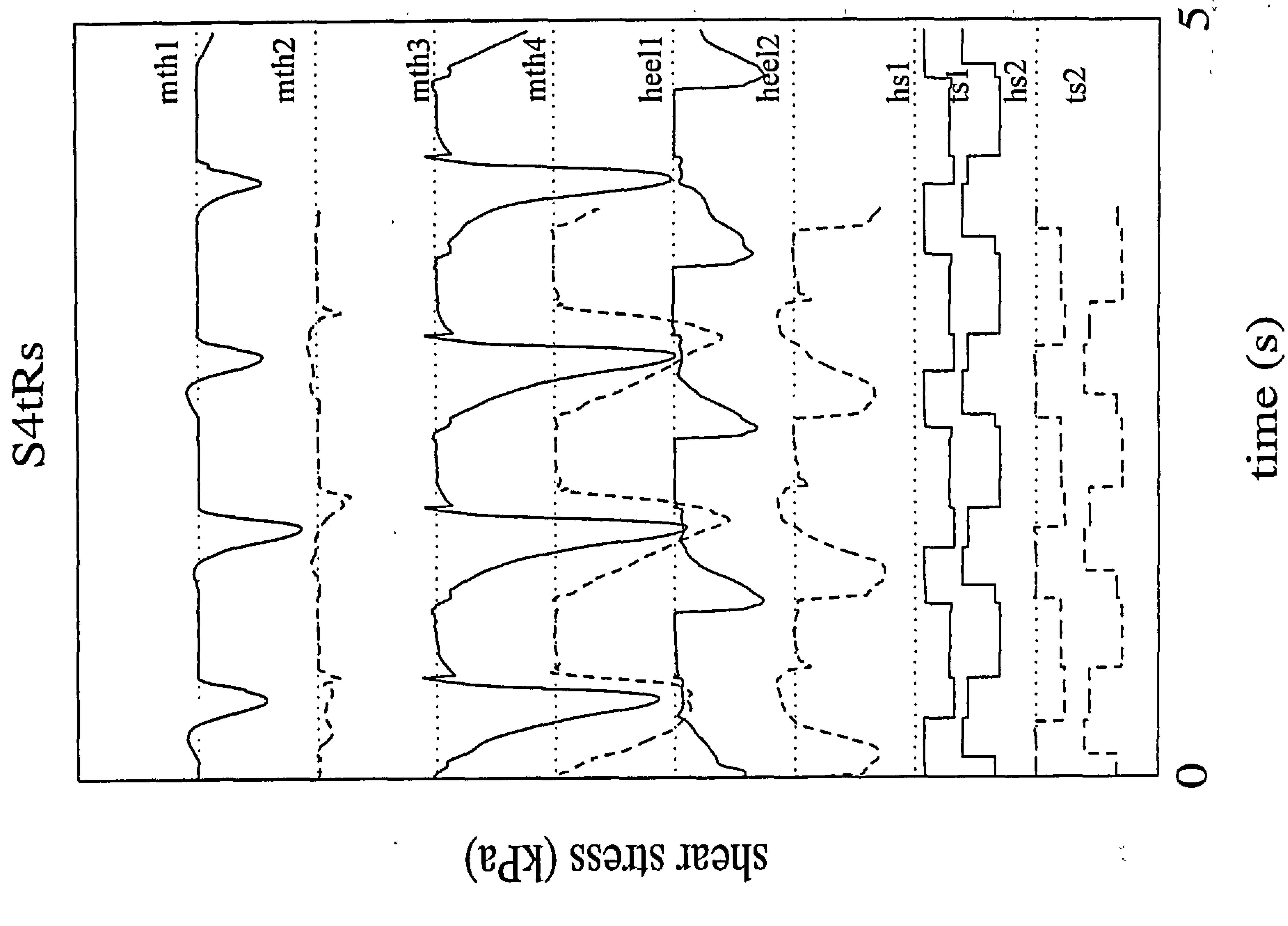

S31Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

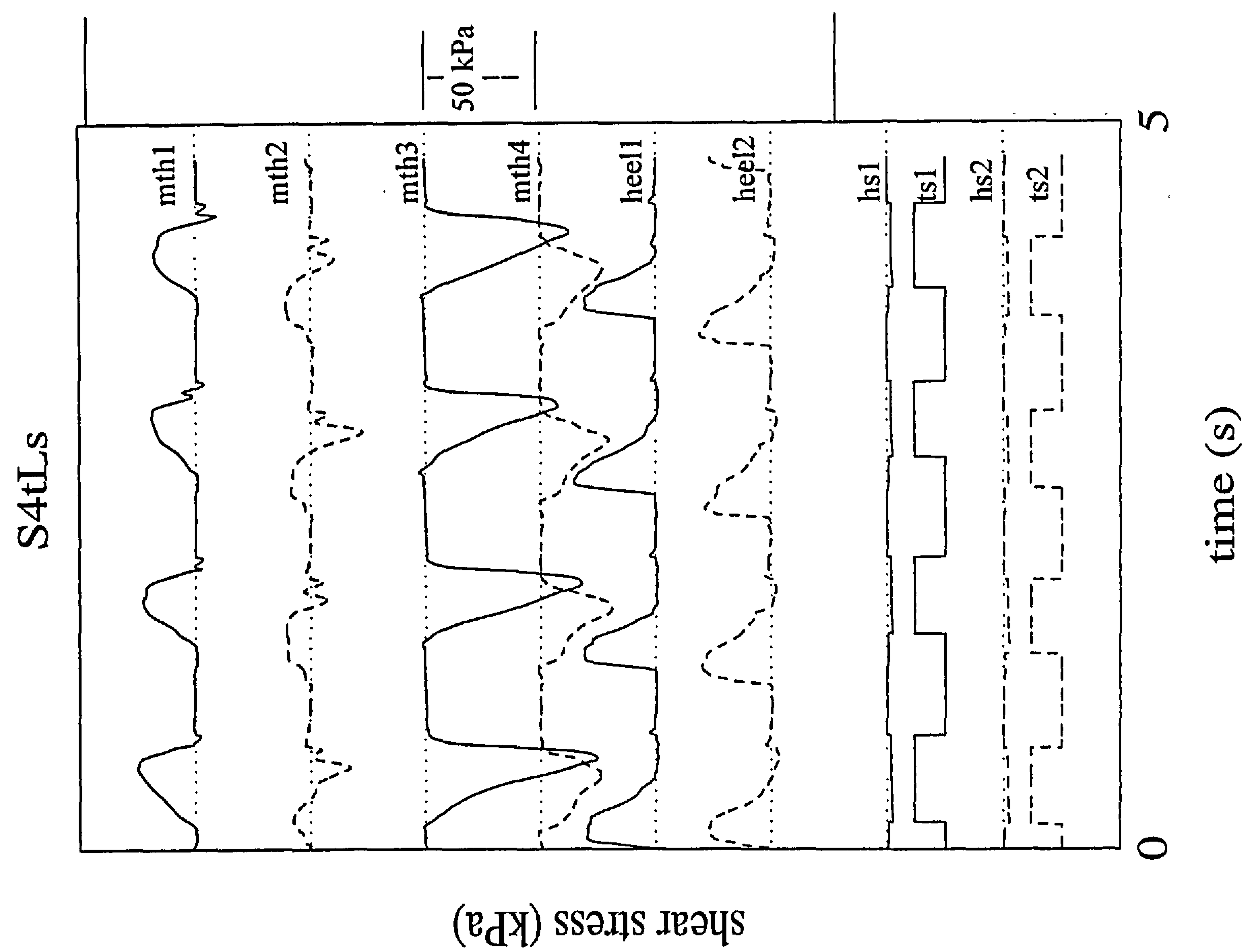

S3tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

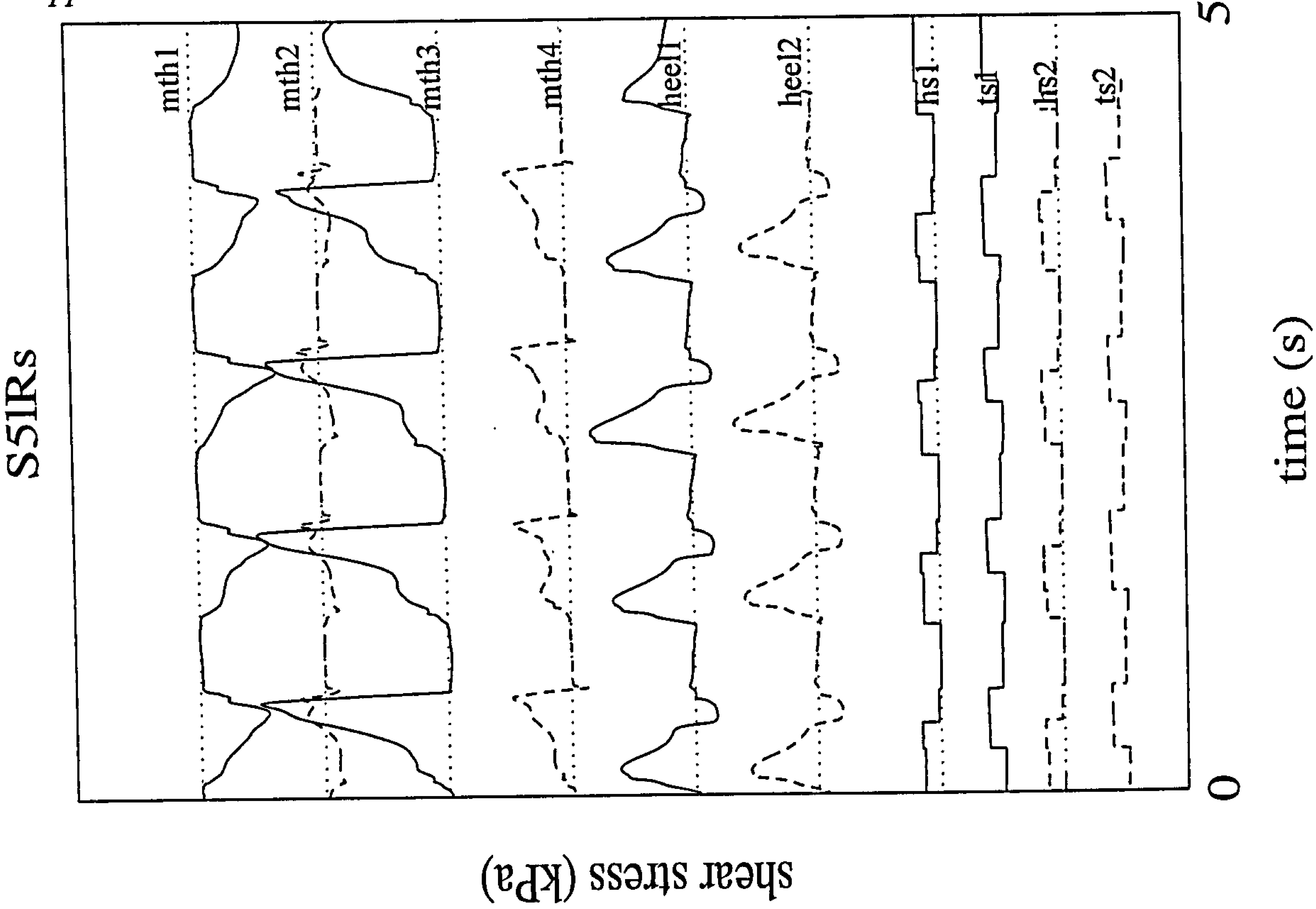

S3tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

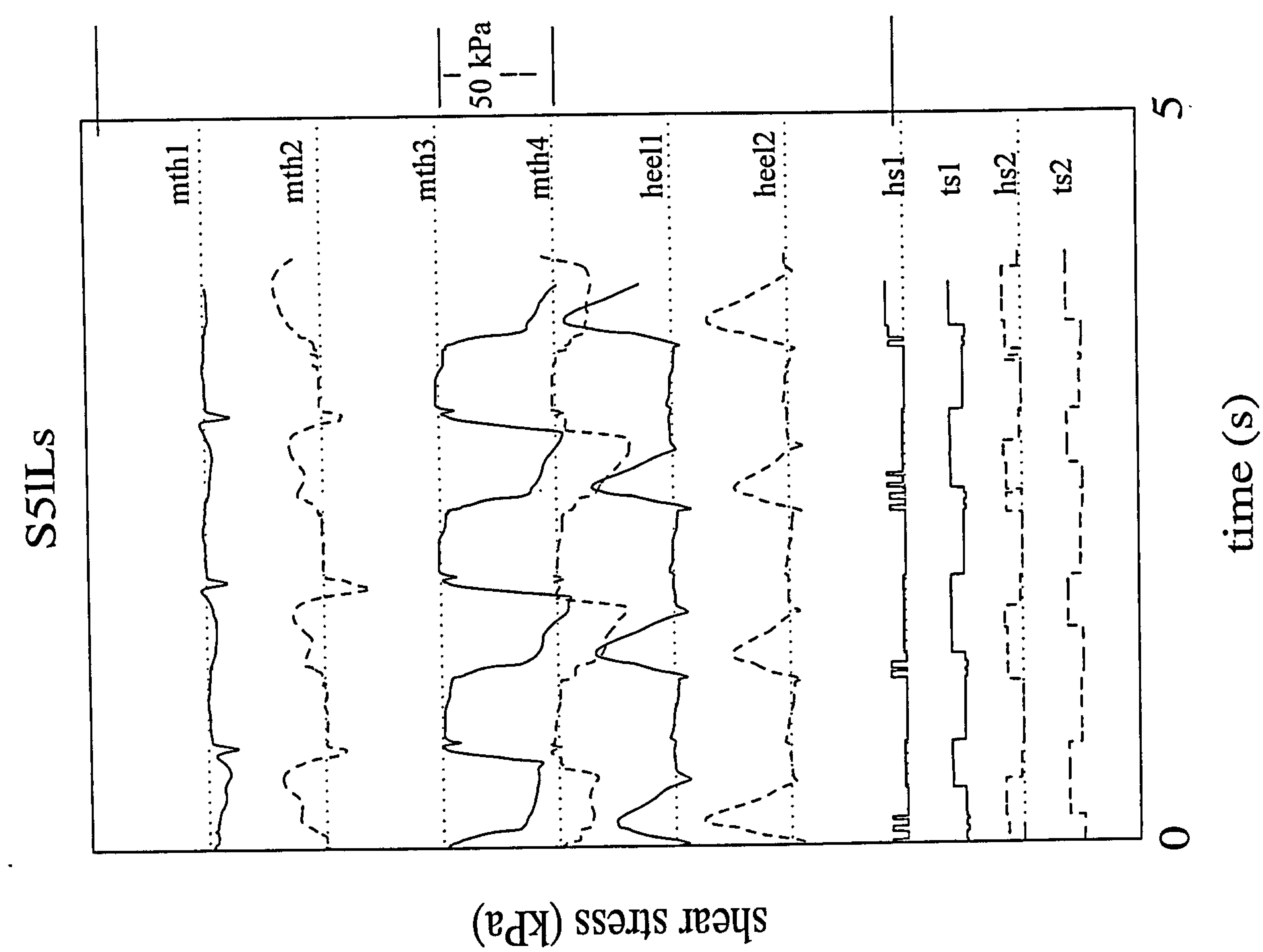

S41Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

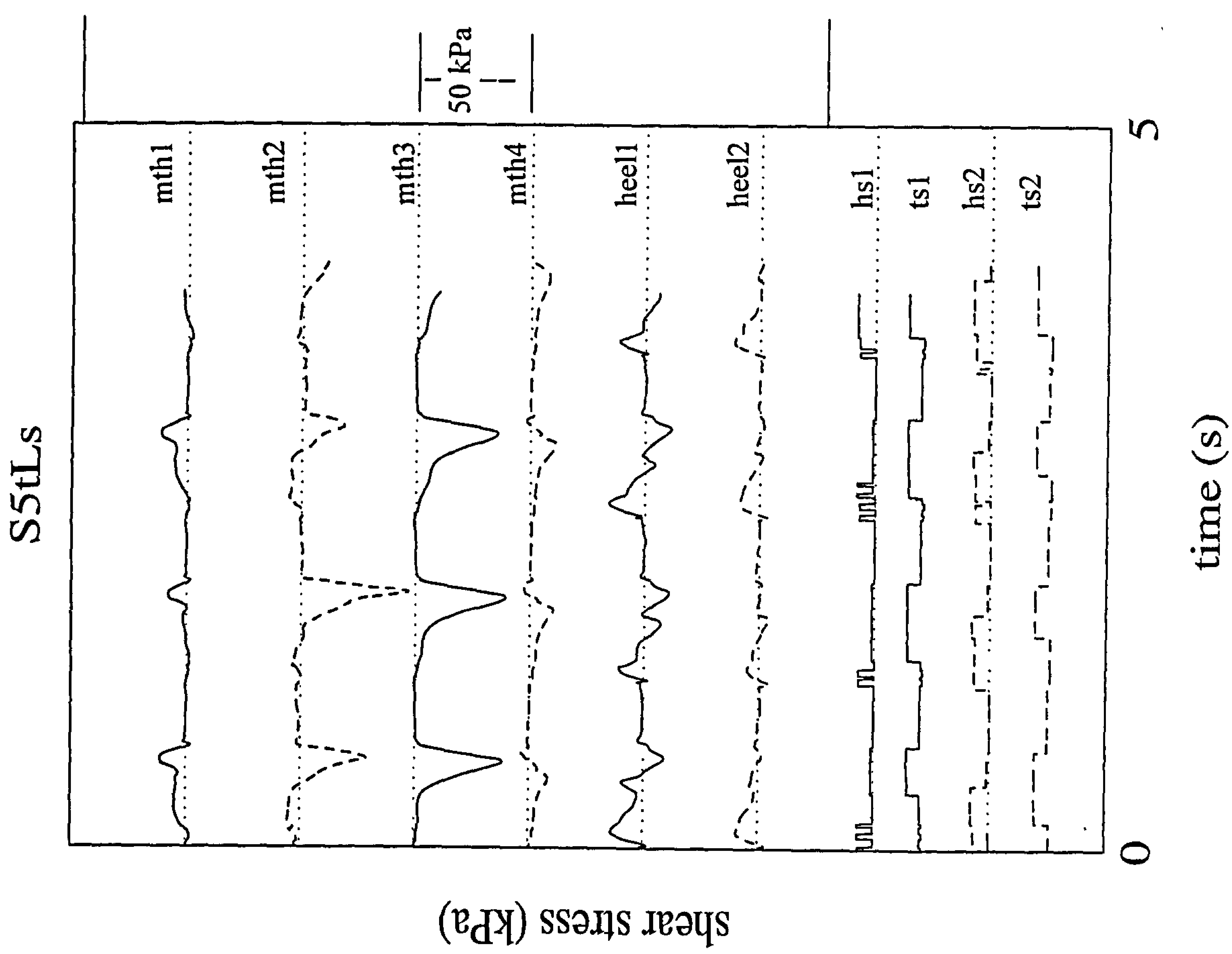

S41Ls
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
time (s)
0
5

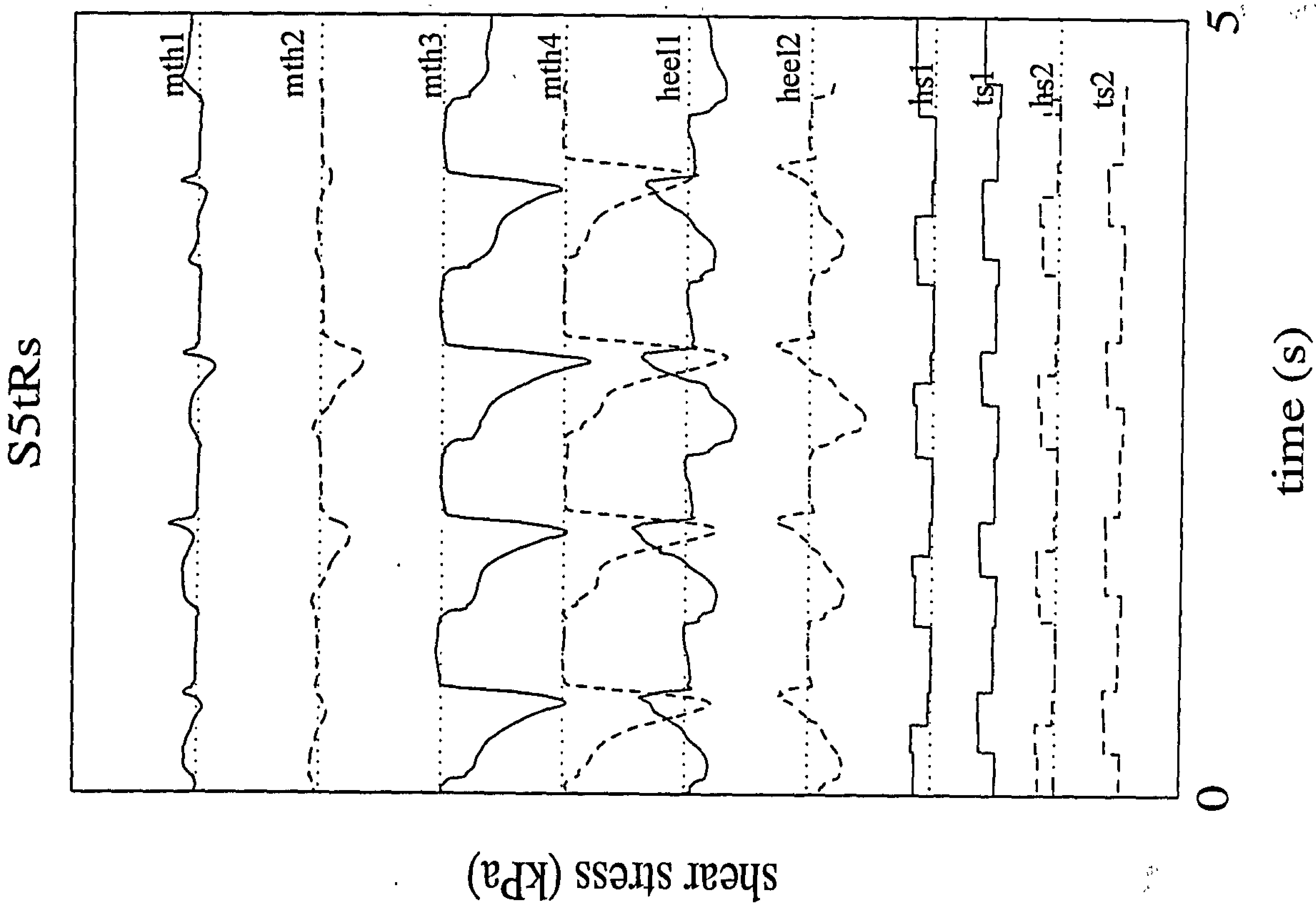

S4tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

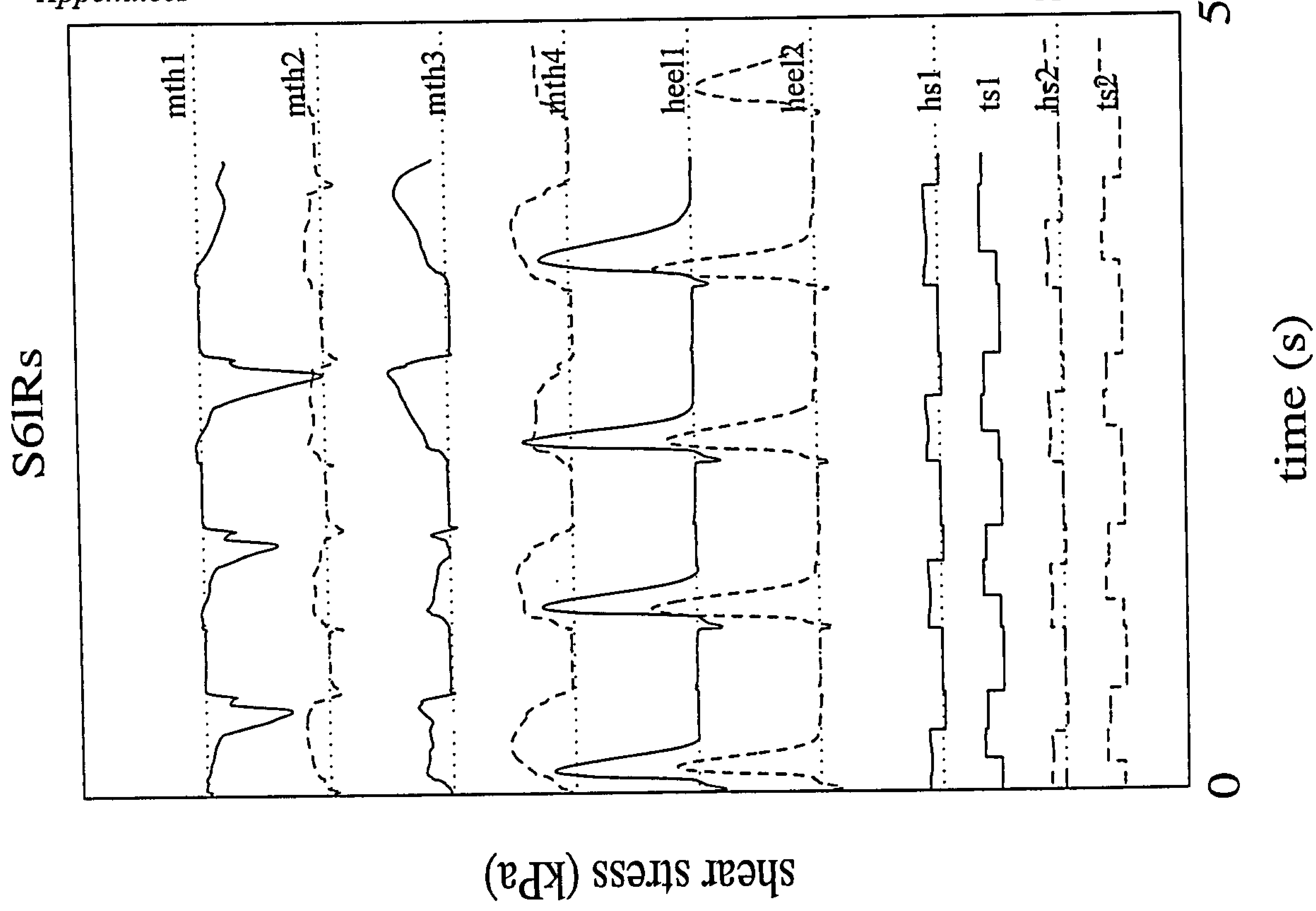

S4tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

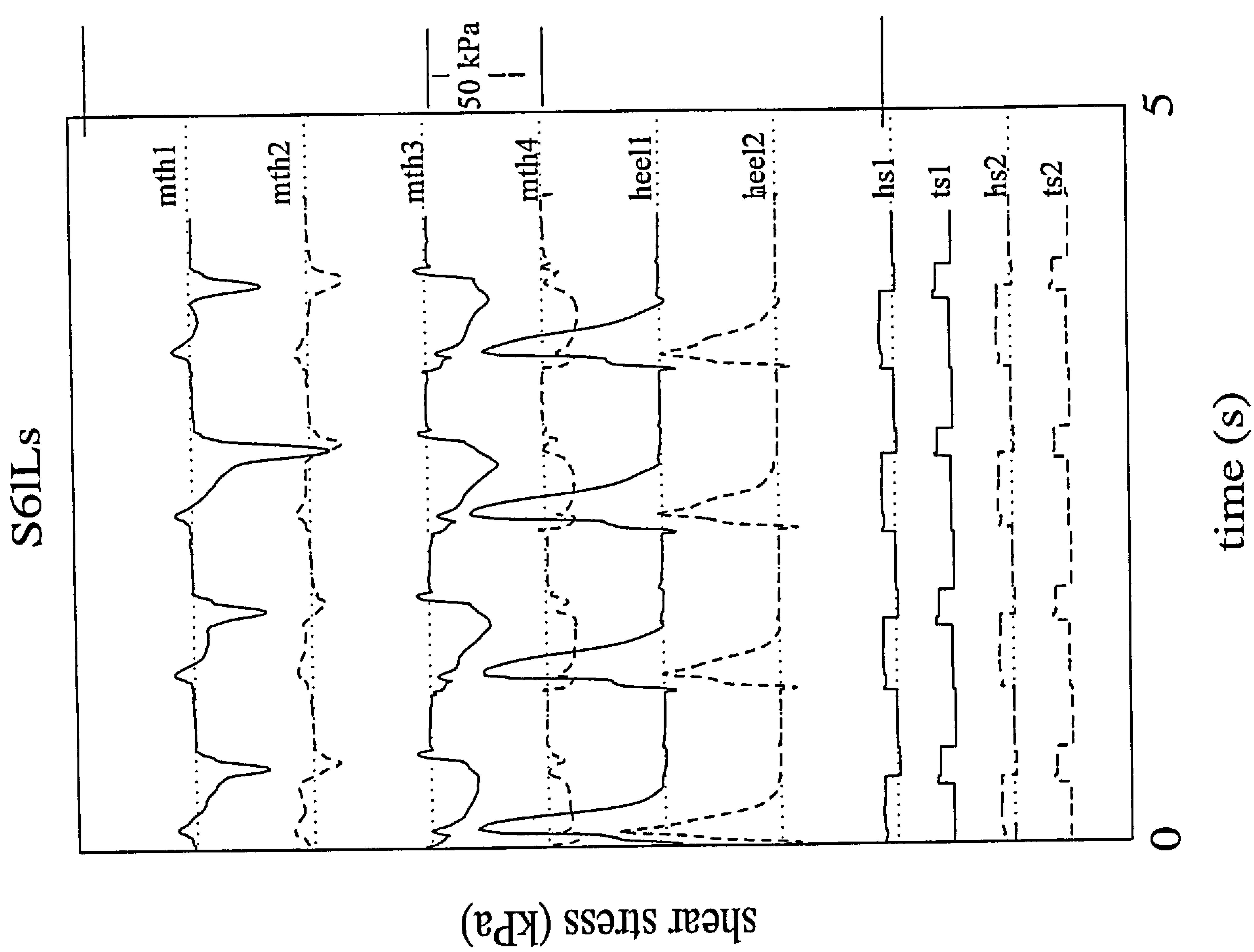

S51Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

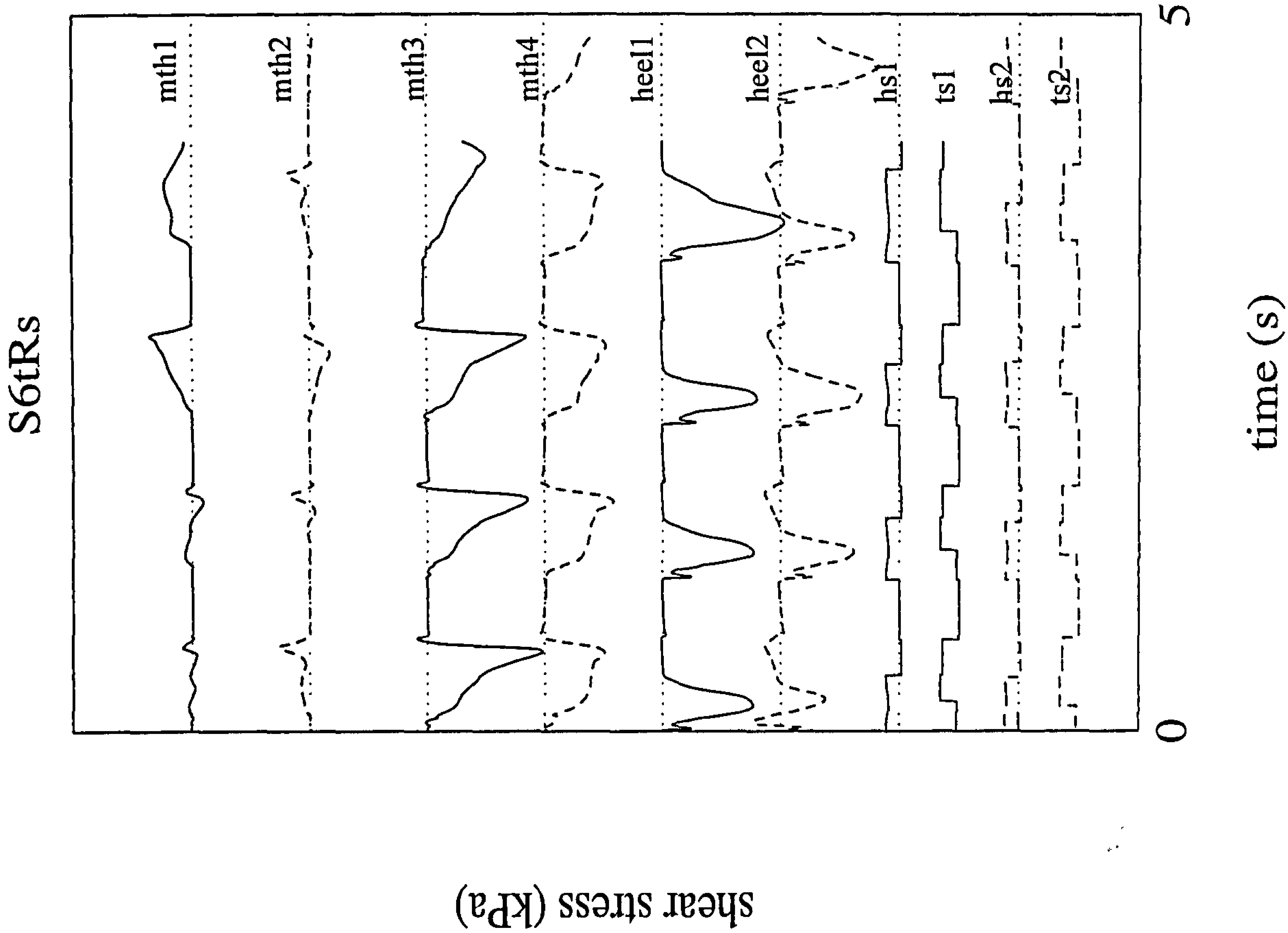

S51Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

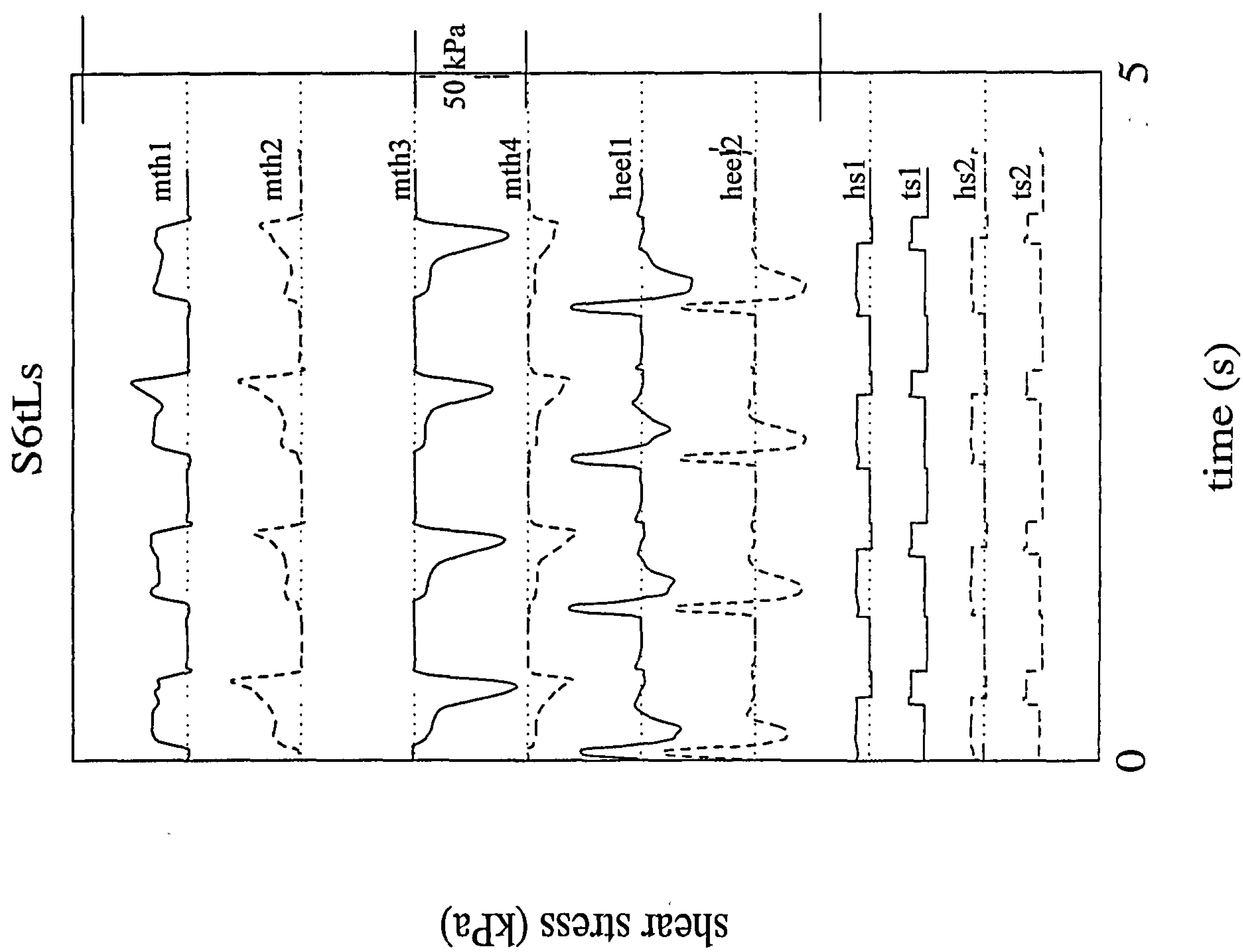

S5tLs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
0
5
time (s)

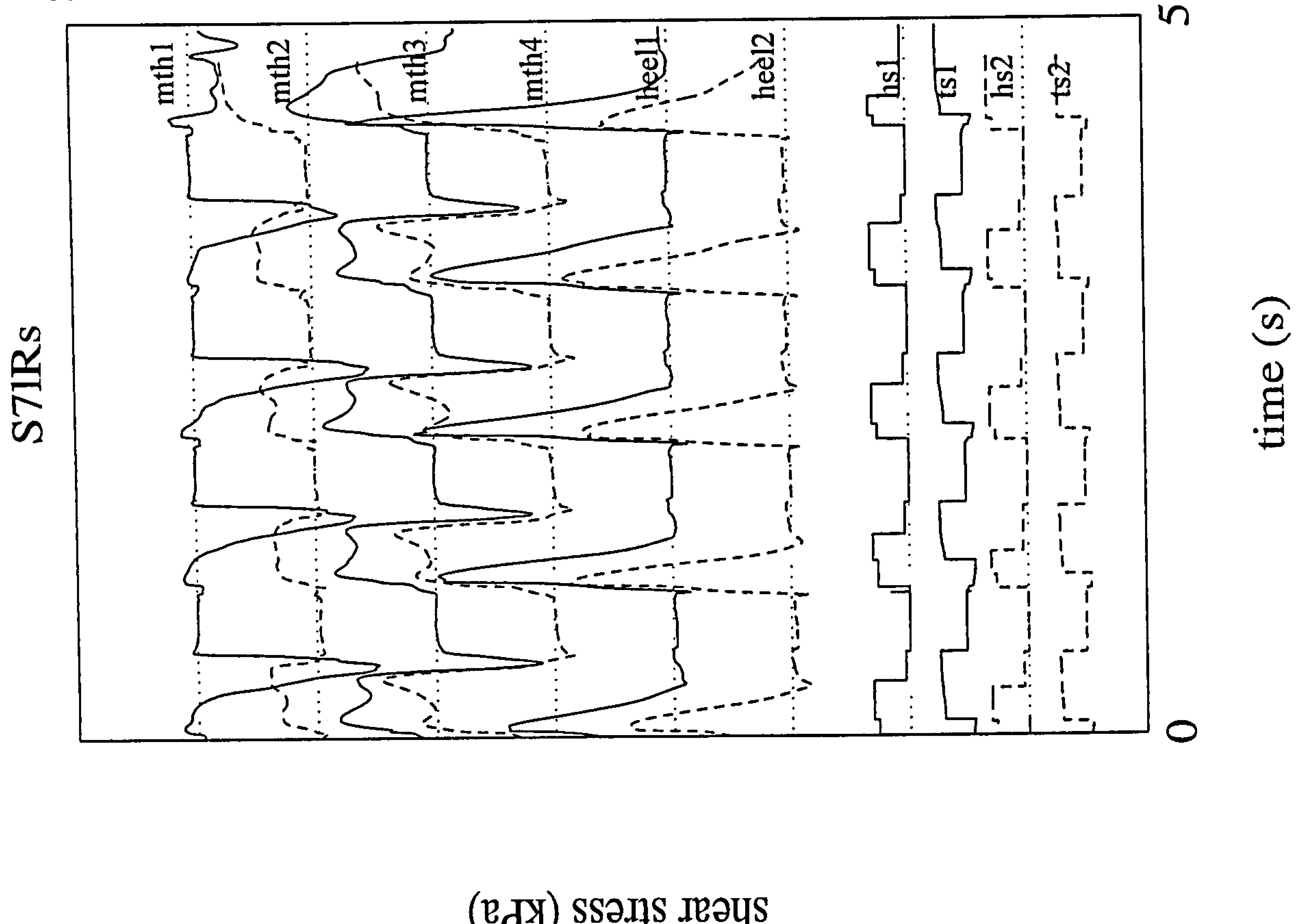

S5tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)

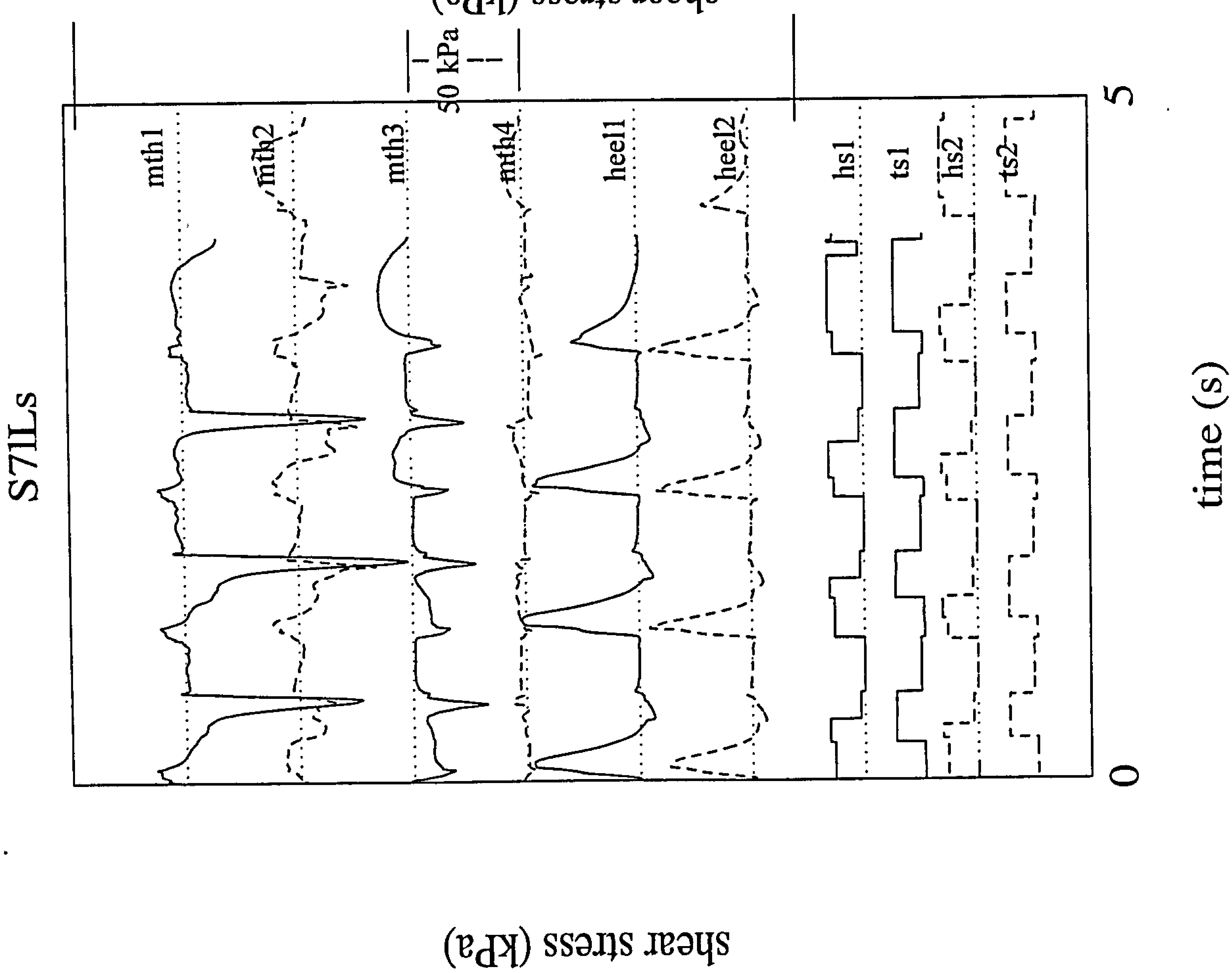

S61Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

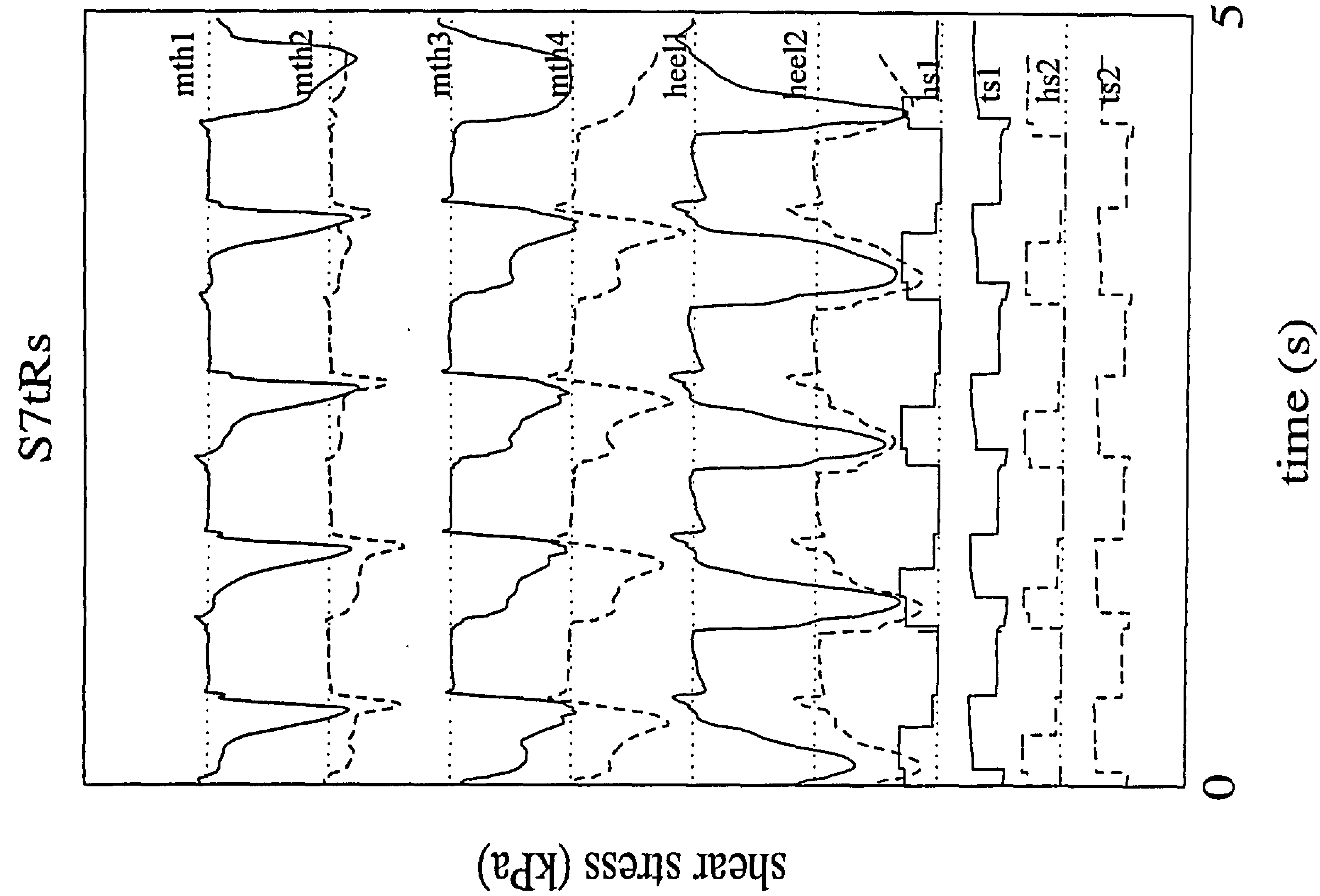

S61Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

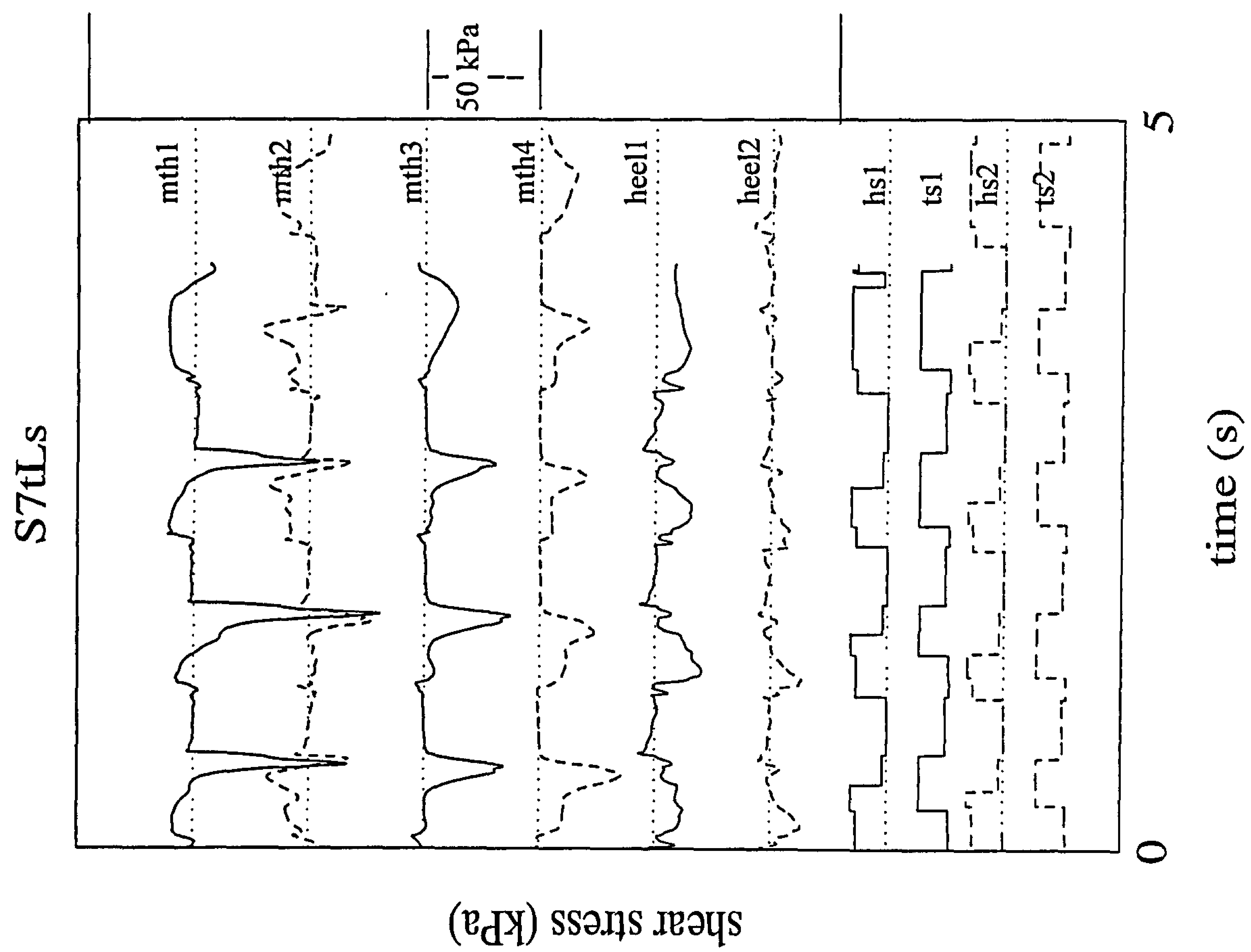

S6tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

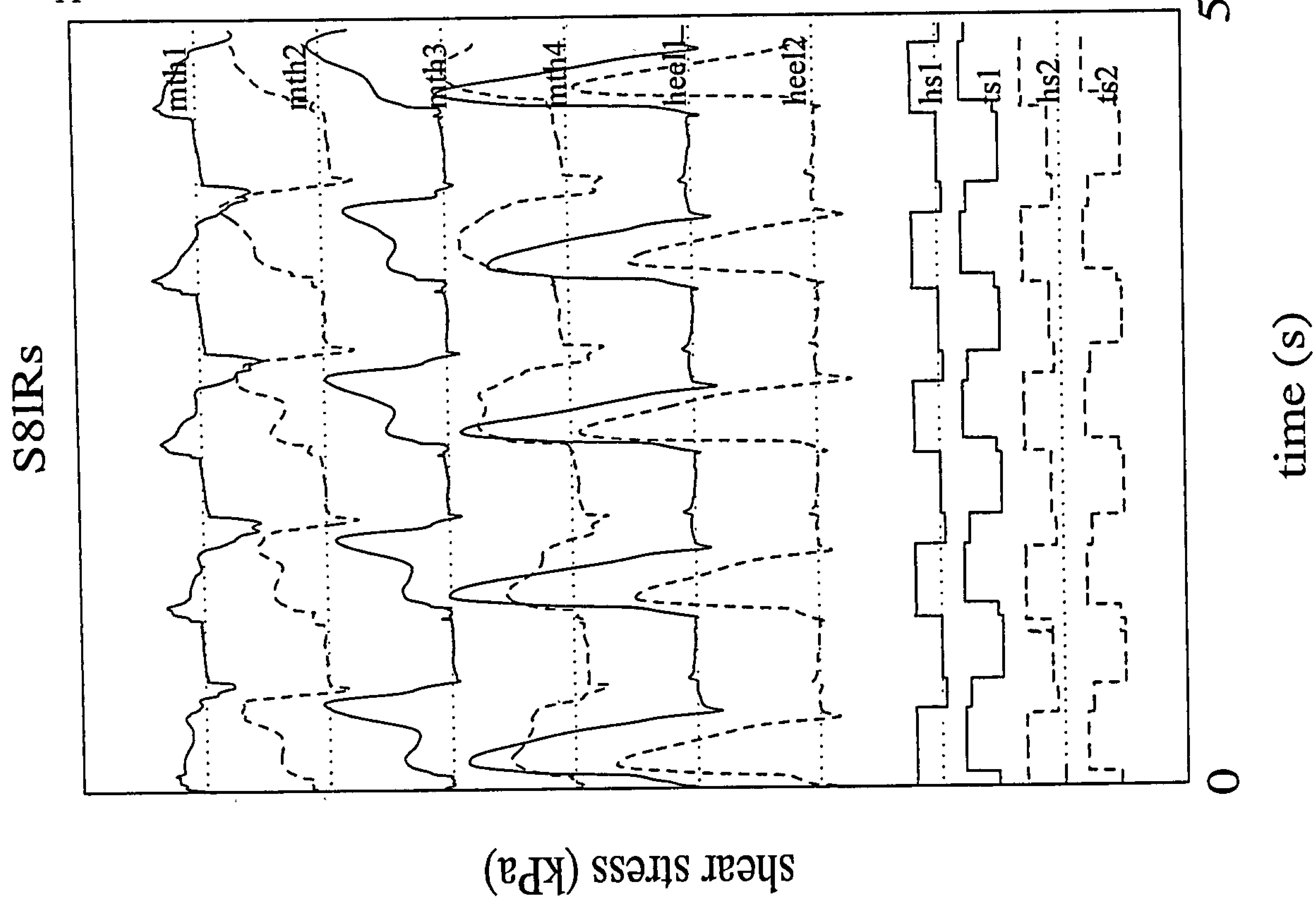

S6tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

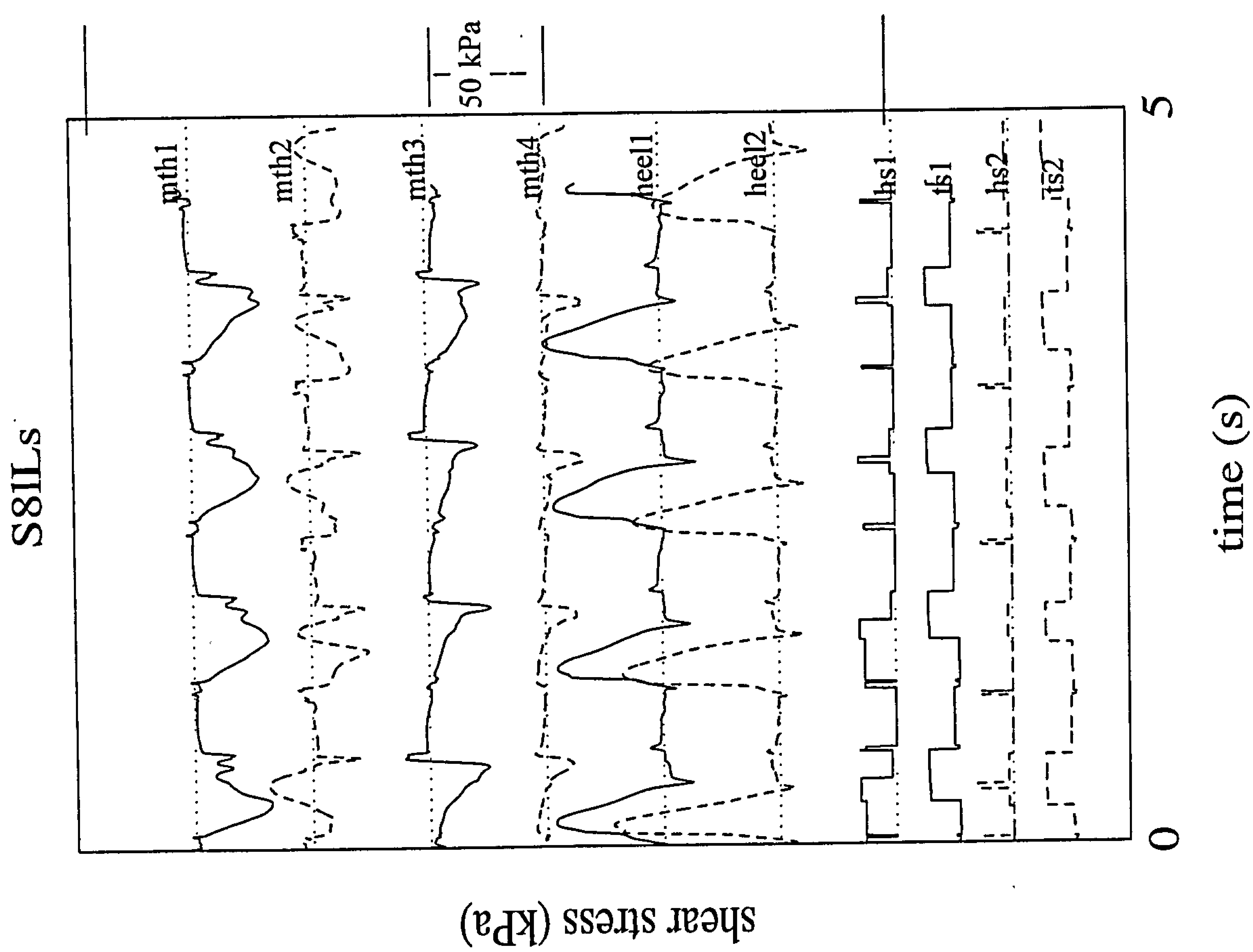

S71Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

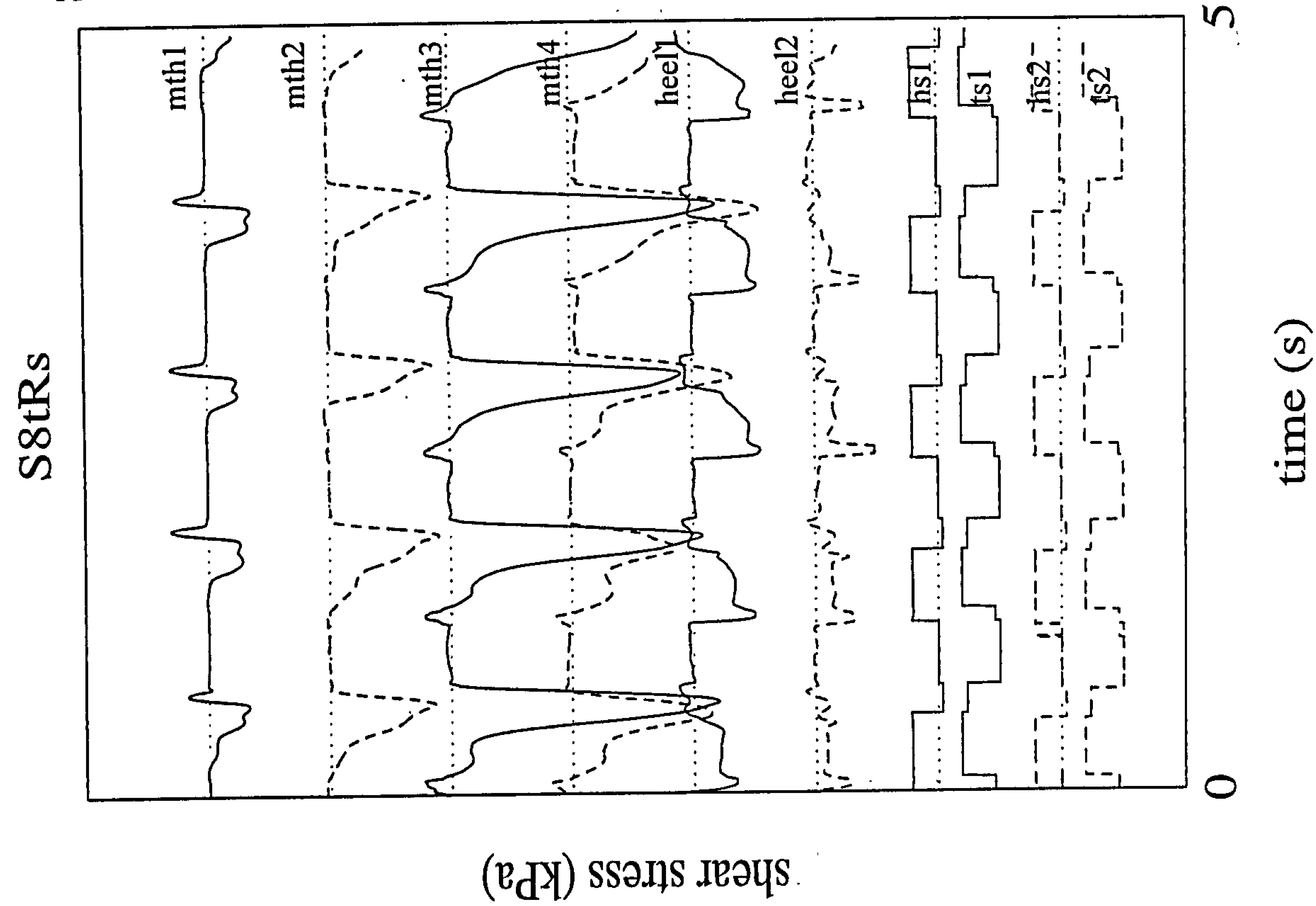

S71Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

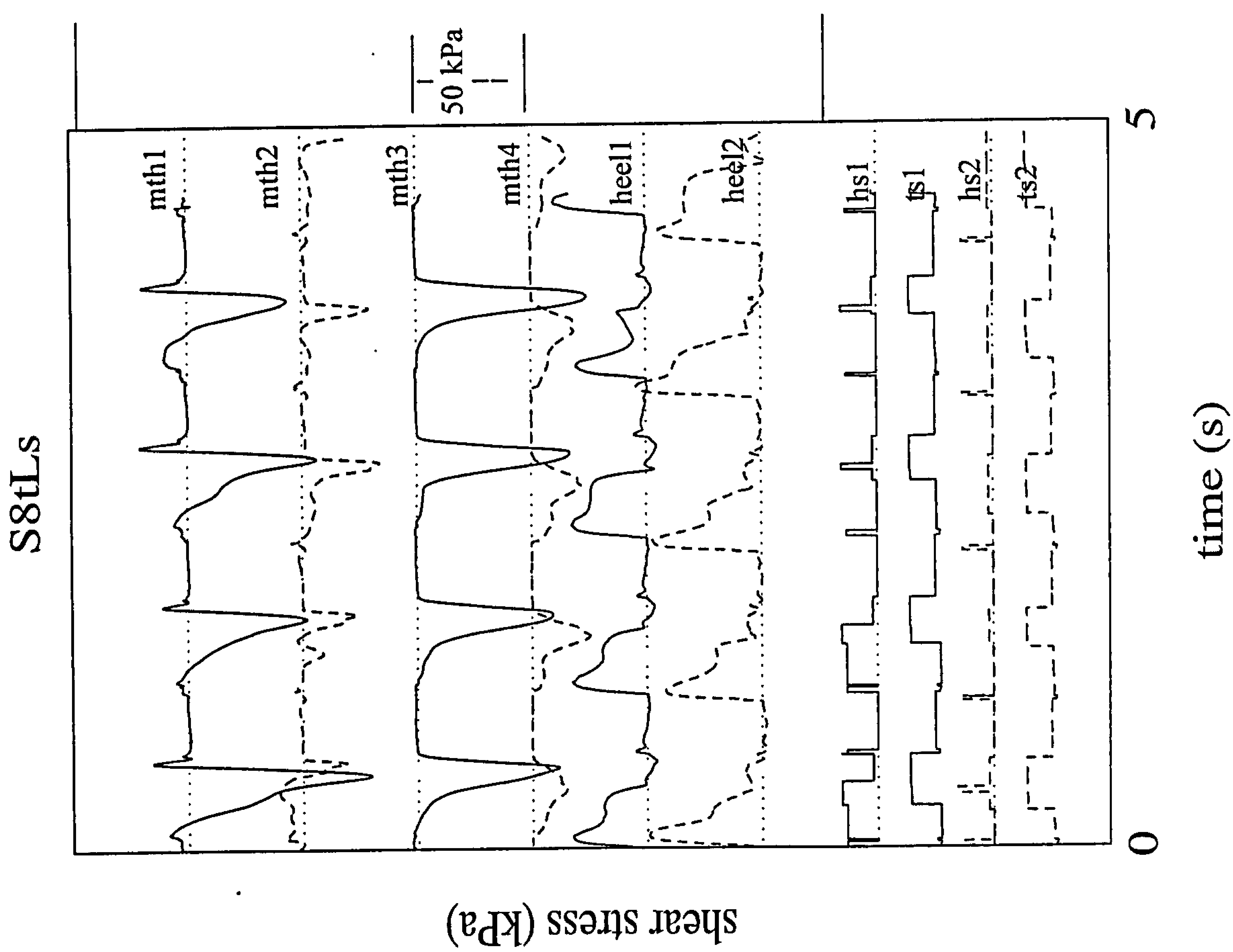

S7tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

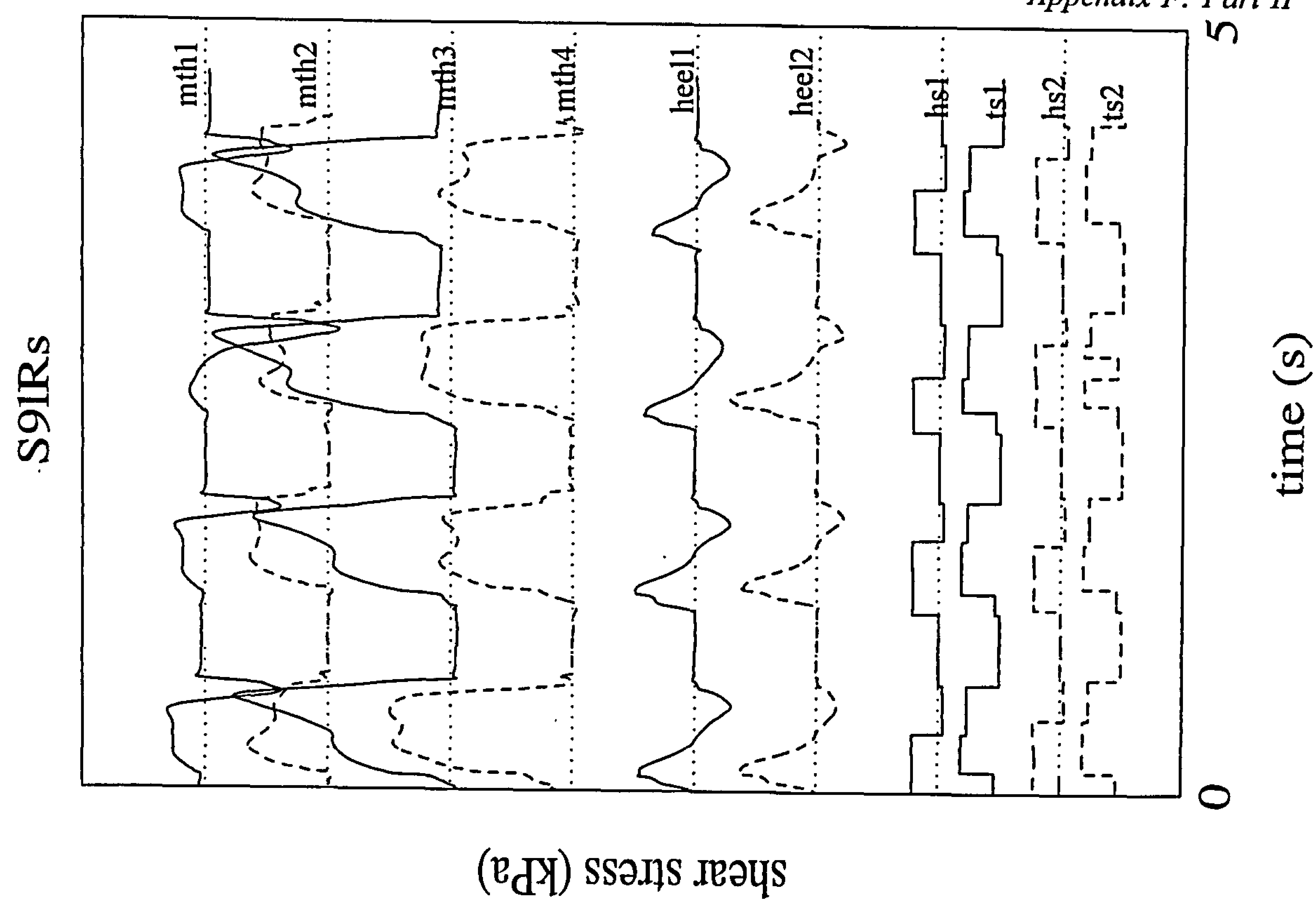

S7tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

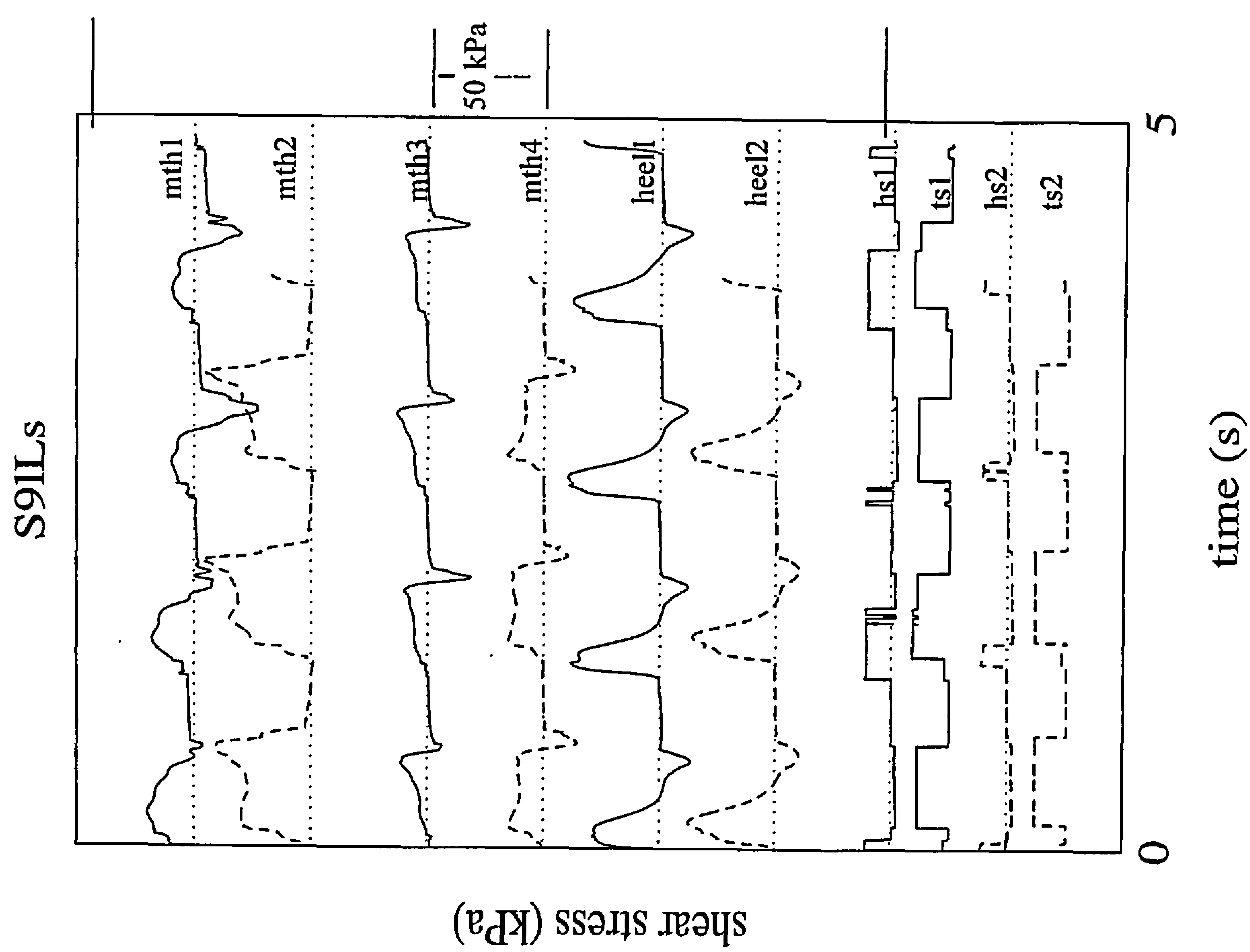

S8IRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

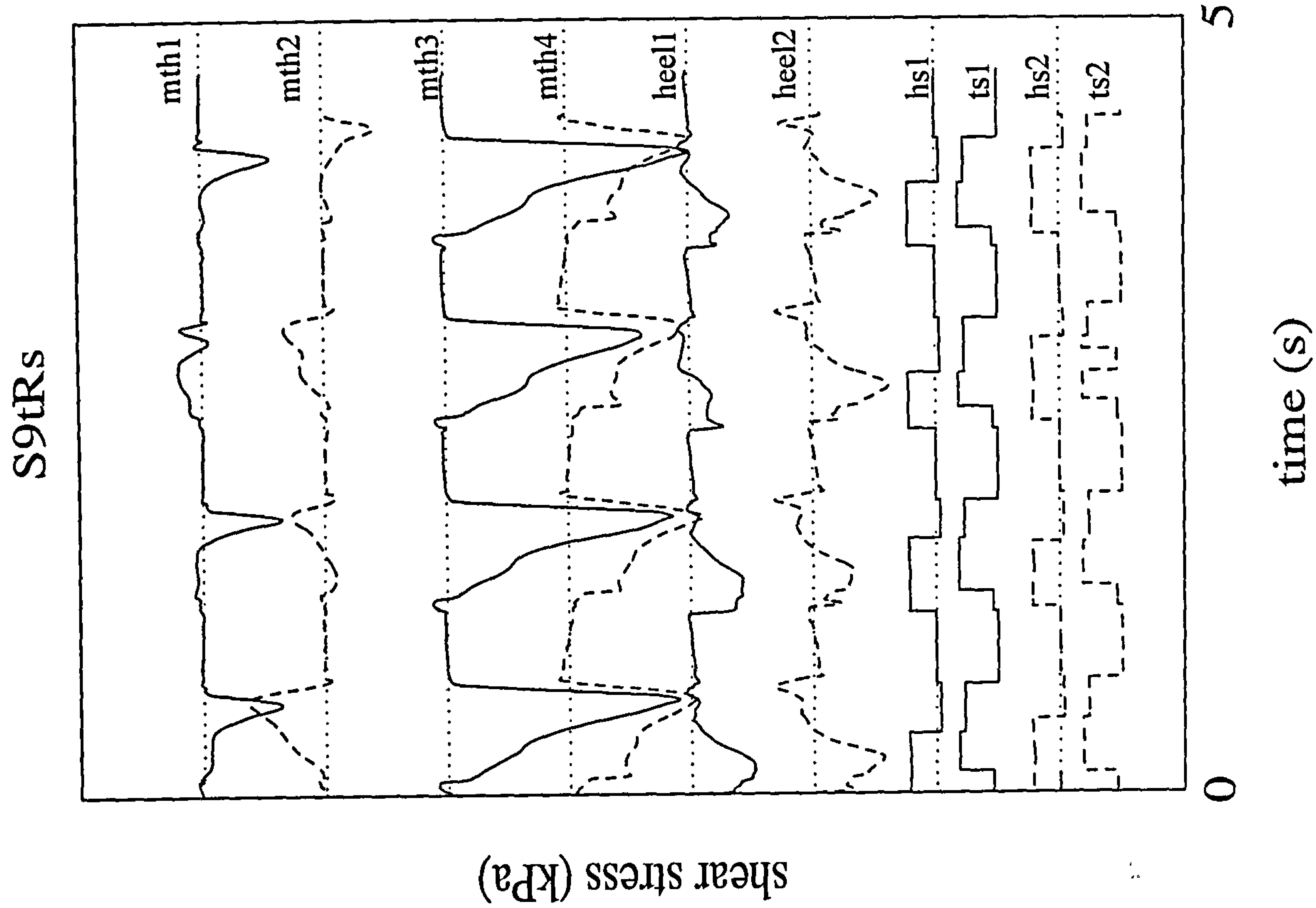

S8ILs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

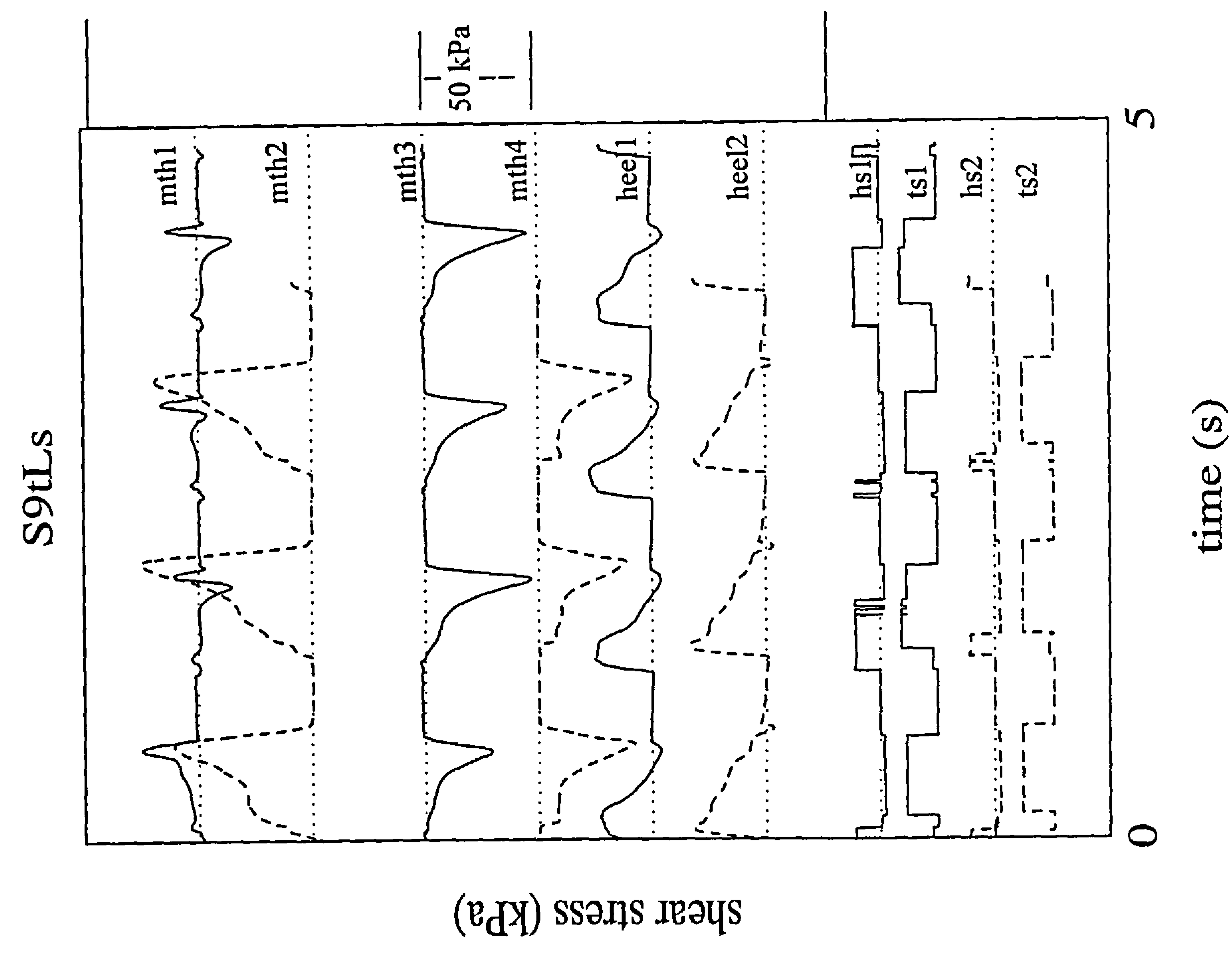

S8tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

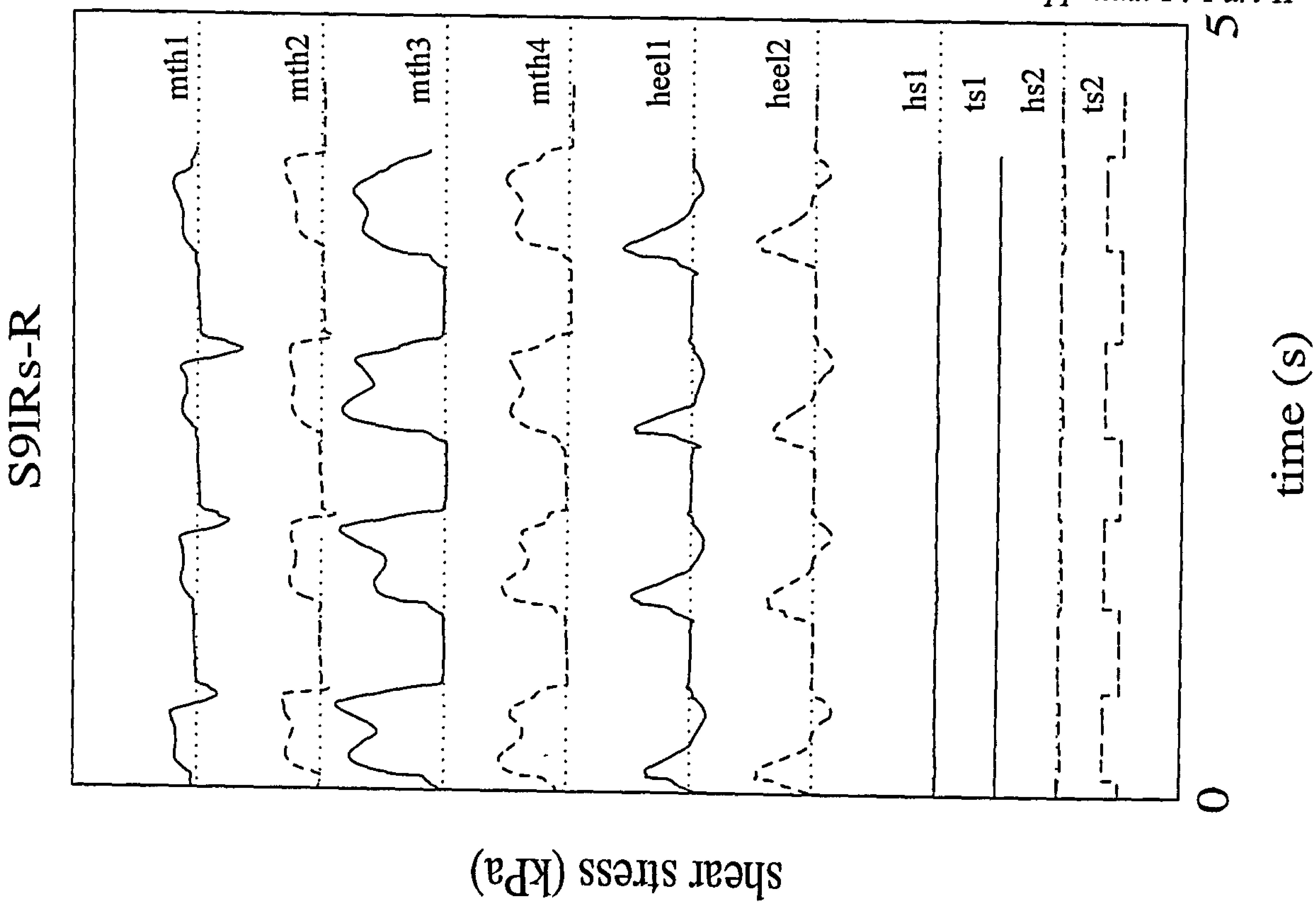

S8tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

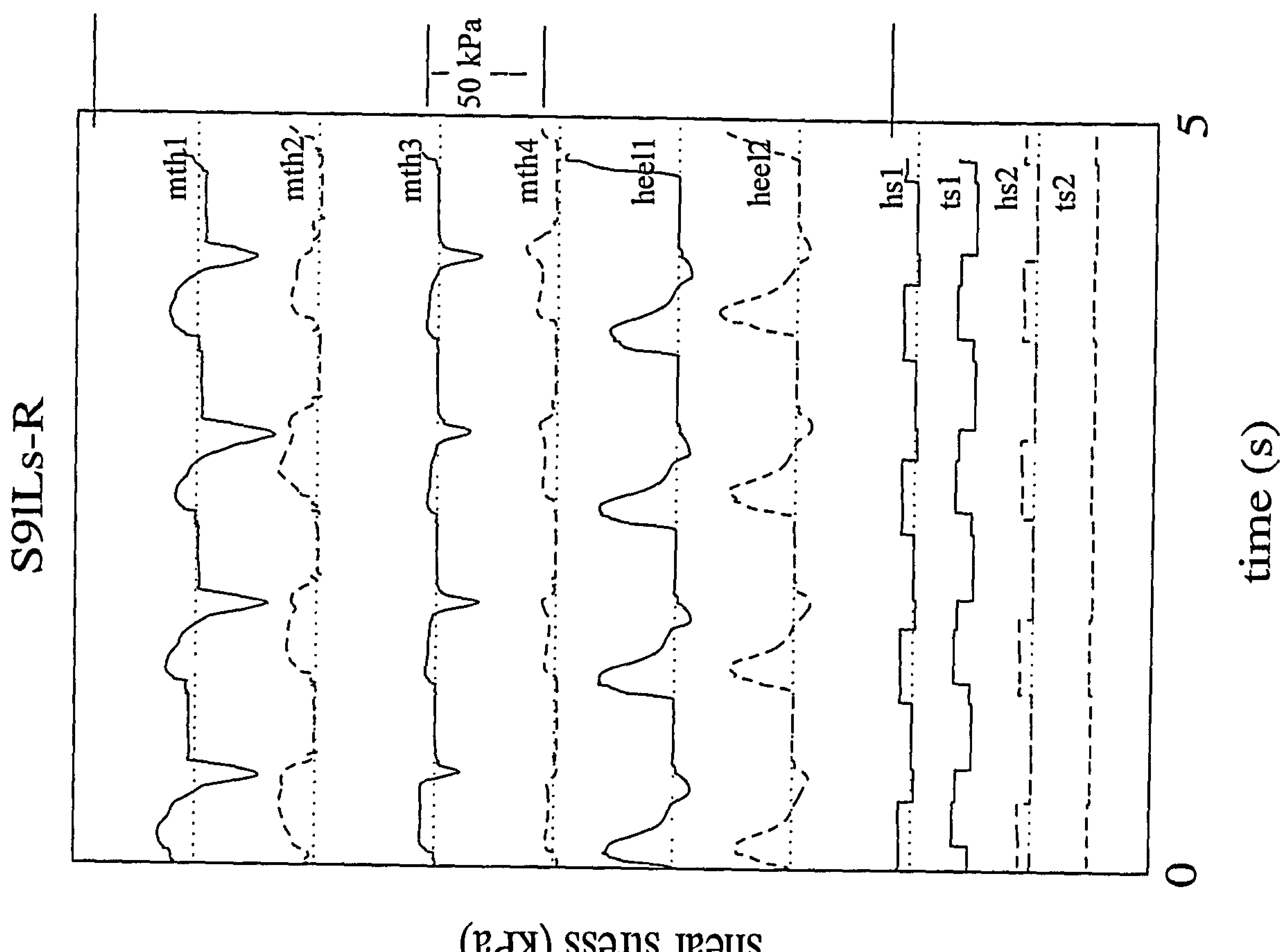

S91Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

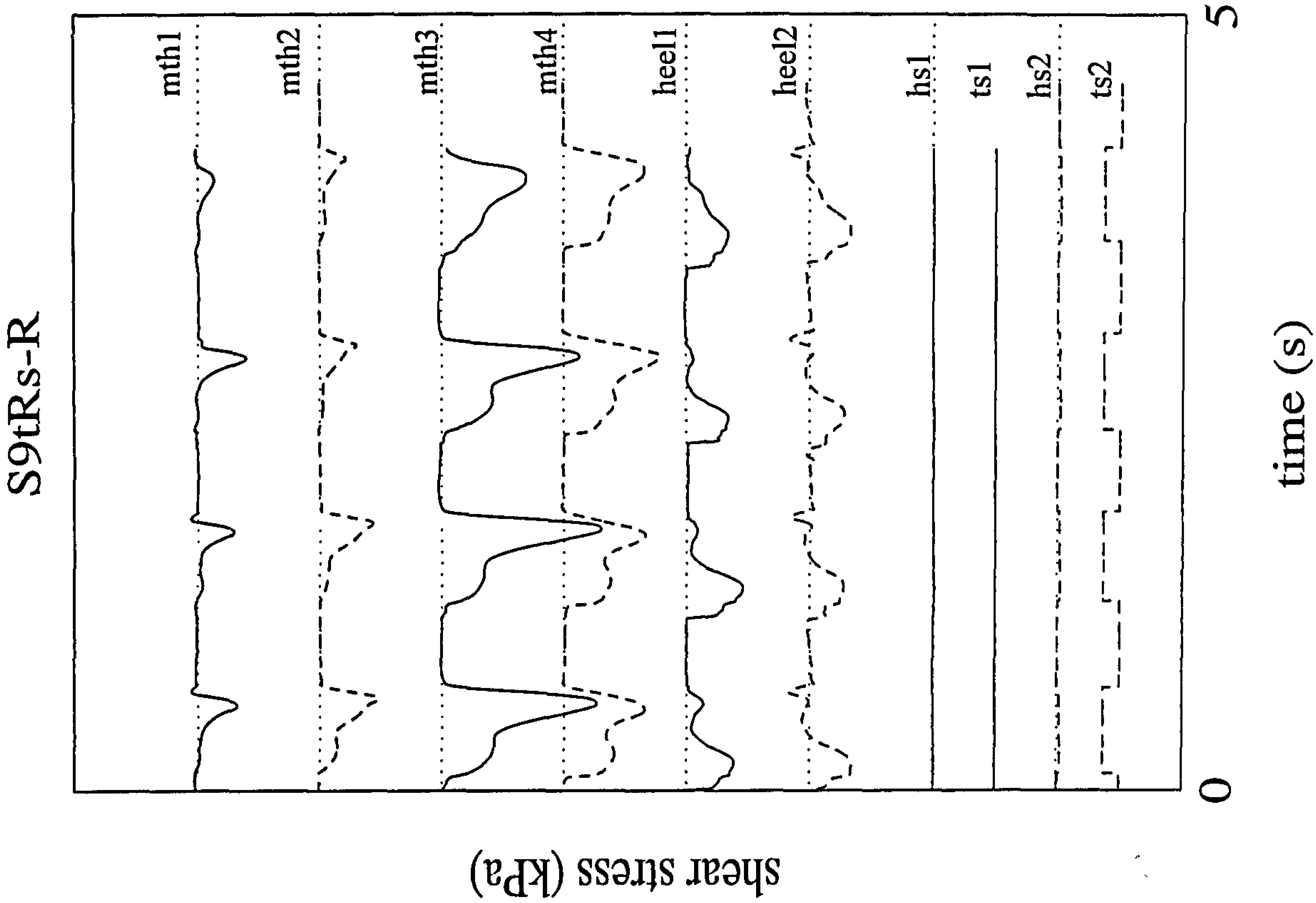

S91Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

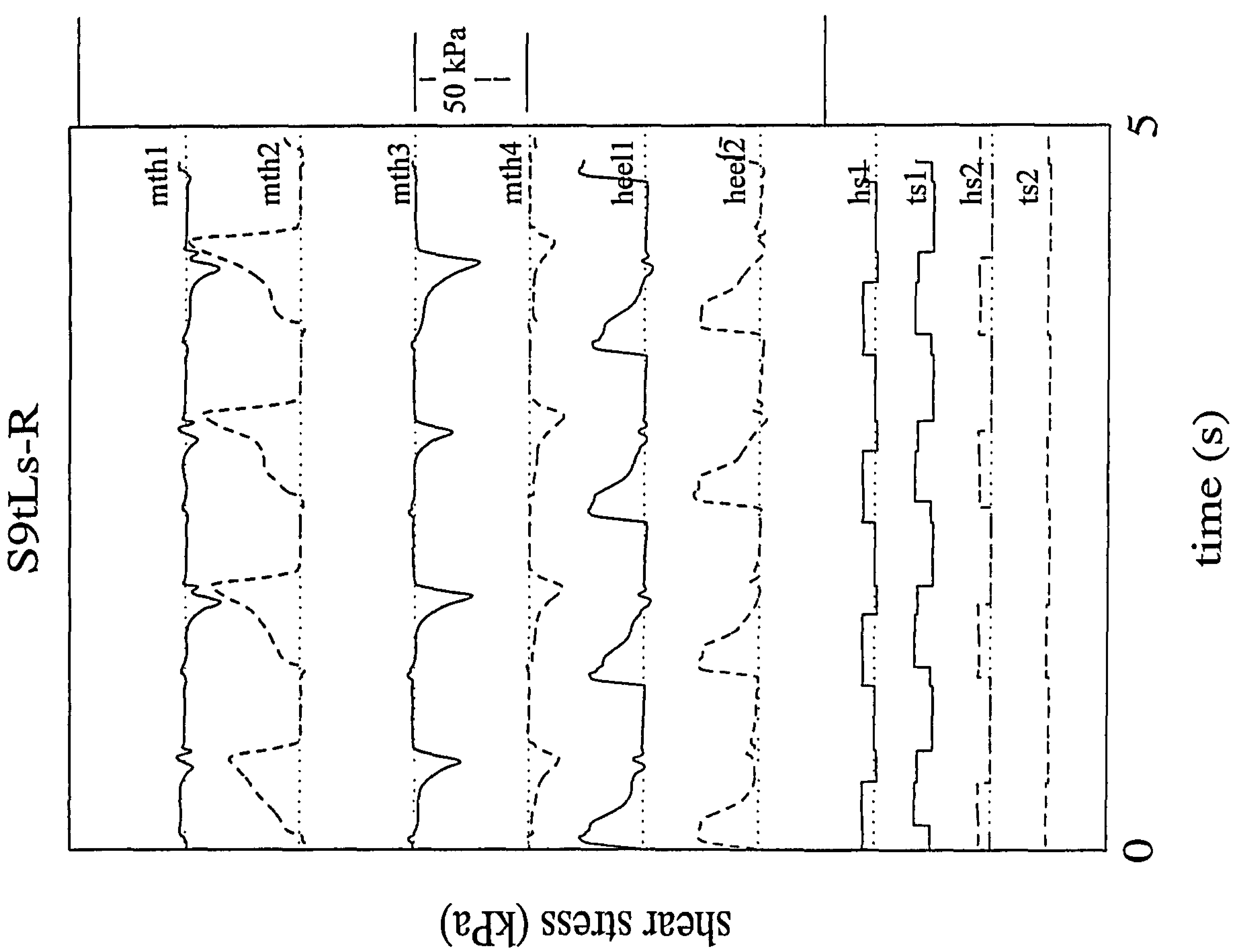

S9tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


S9tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


S91Rs-R
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


S91Ls-R
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


S9tRs-R
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


S9tLs-R
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

APPENDIX G

coding system for the shear records

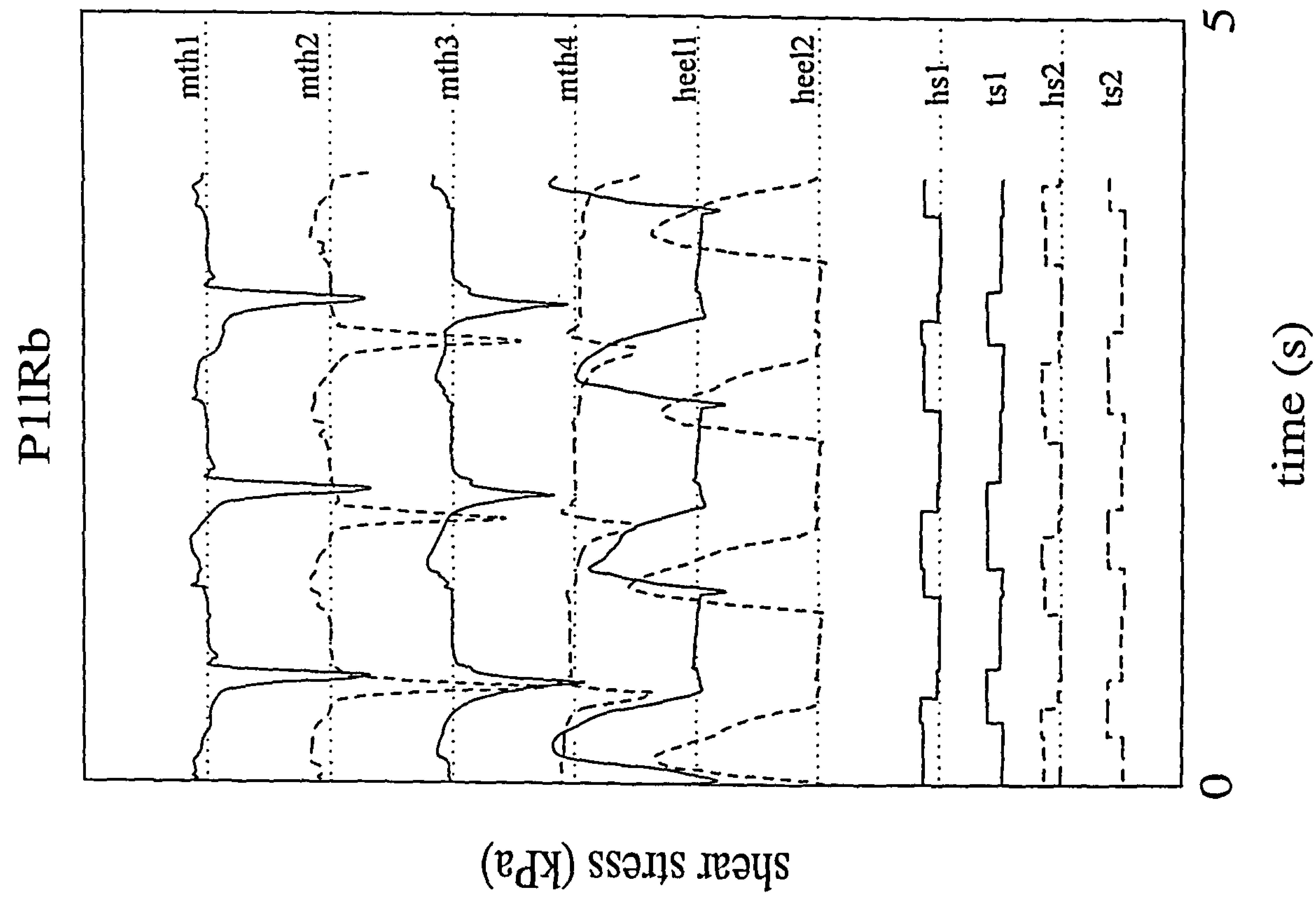
The shear records in this Appendix are titled using a simple five character code:

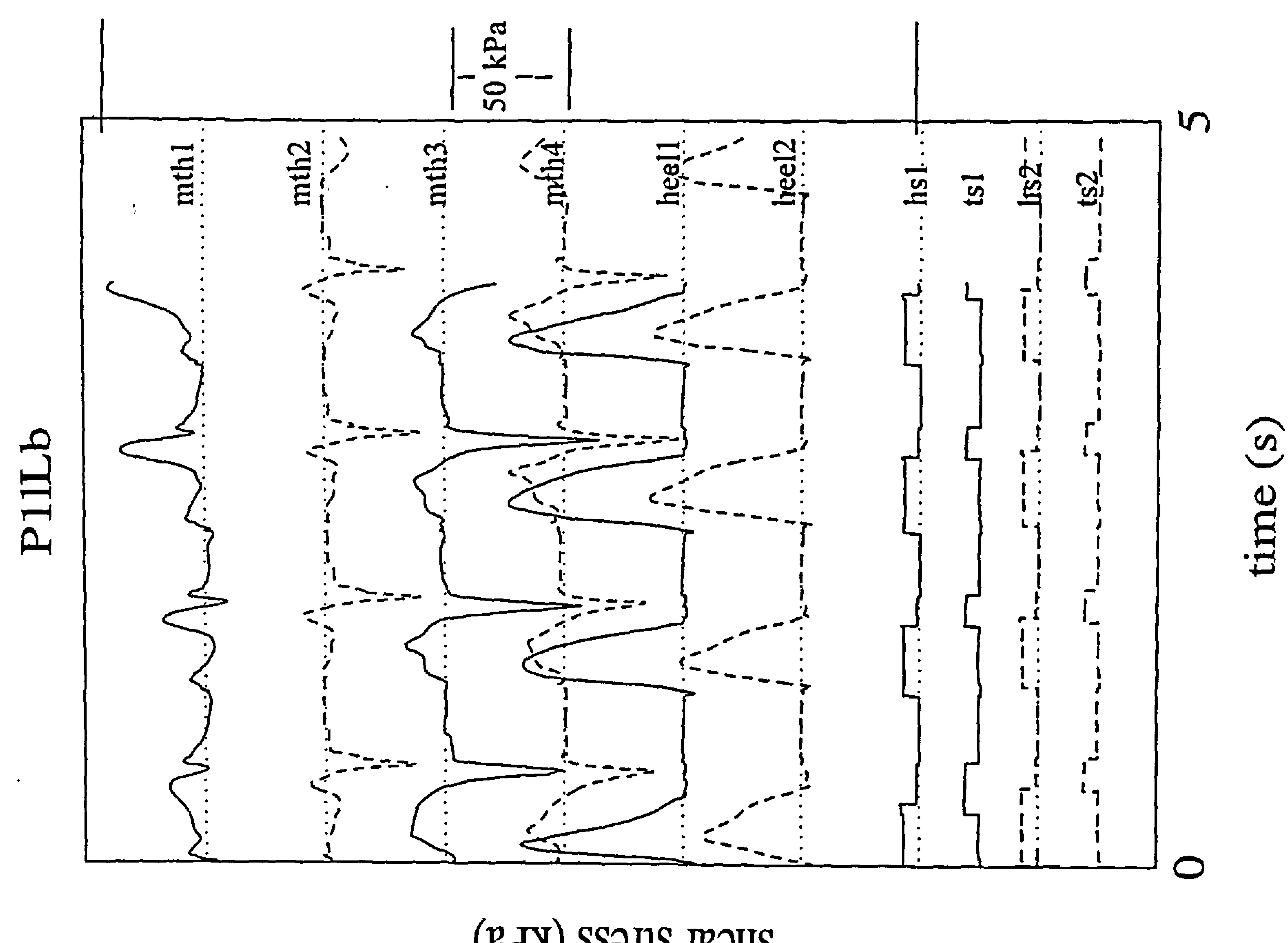
- *P* refers to a Patient shear record;
- a number refers to the particular patient (see table 5.2);
- *l* or *t* refers to a record of longitudinal or transverse shear, respectively;
- *R* or *L* refers to the Right or Left foot, respectively;

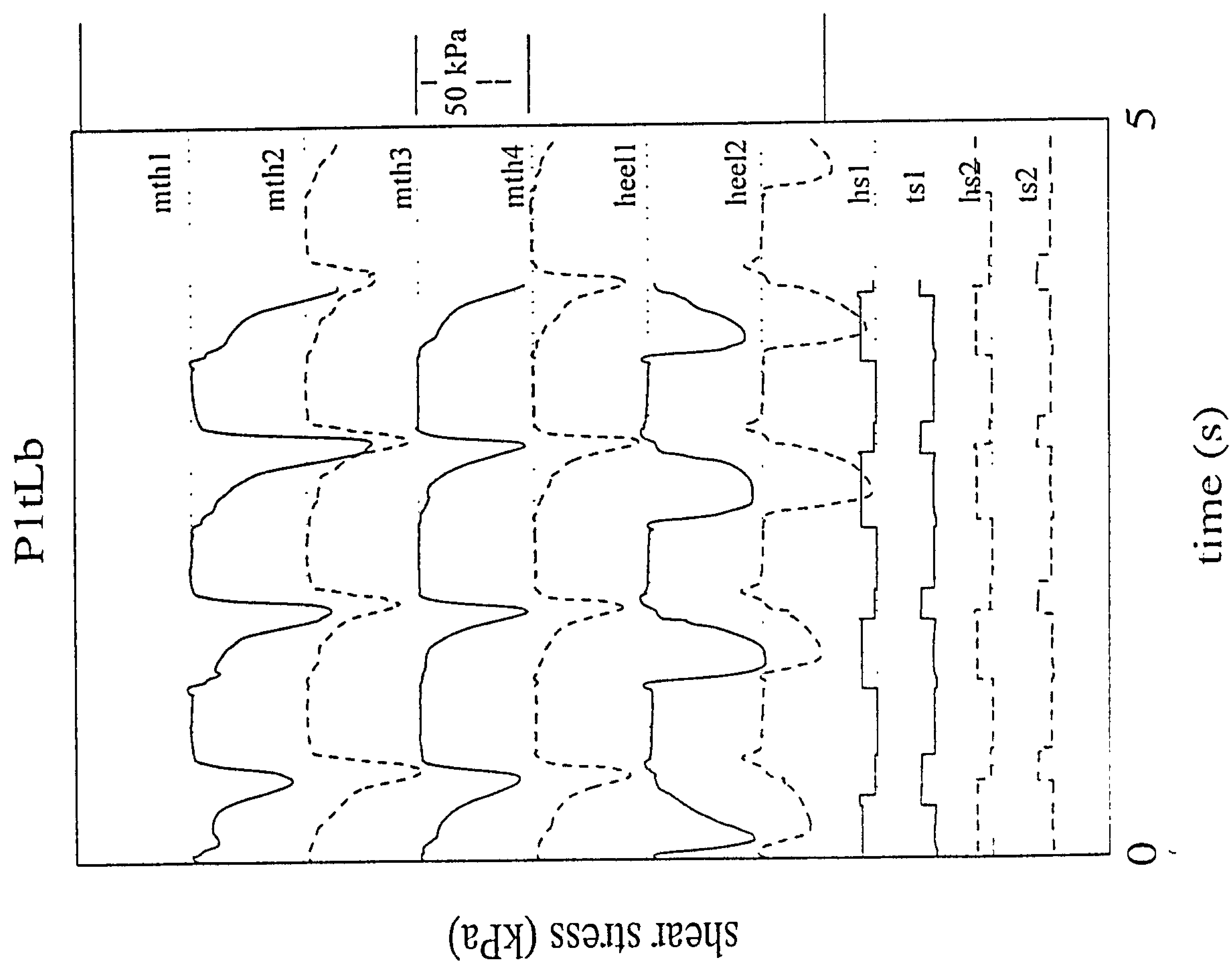
and

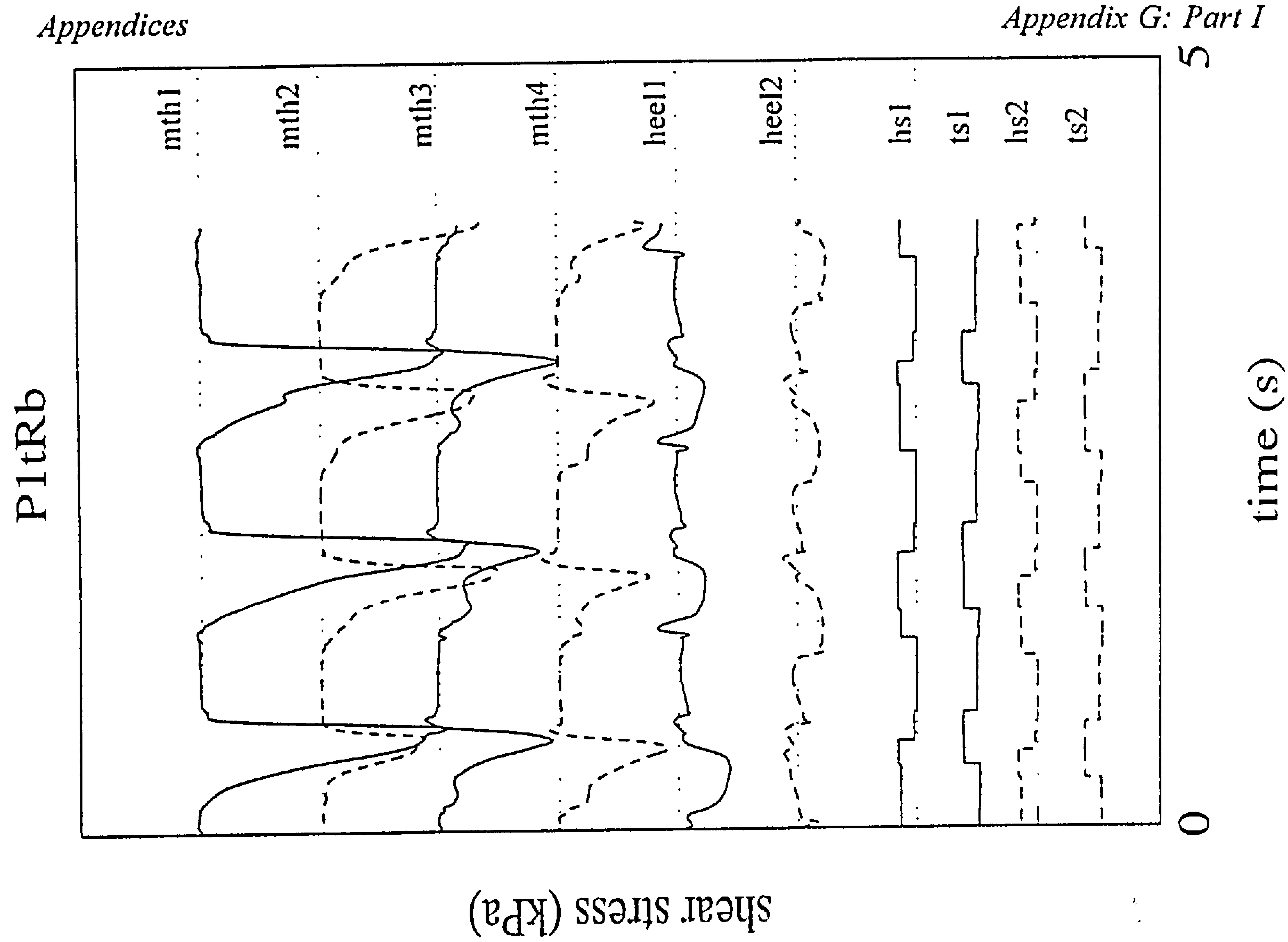
- *s* or *b* refers to the condition under which the recording was made, either with the individual wearing or not wearing nylon hold-ups, respectively, while in the extra-depth stock orthopaedic shoes ie. either wearing socks or barefoot.

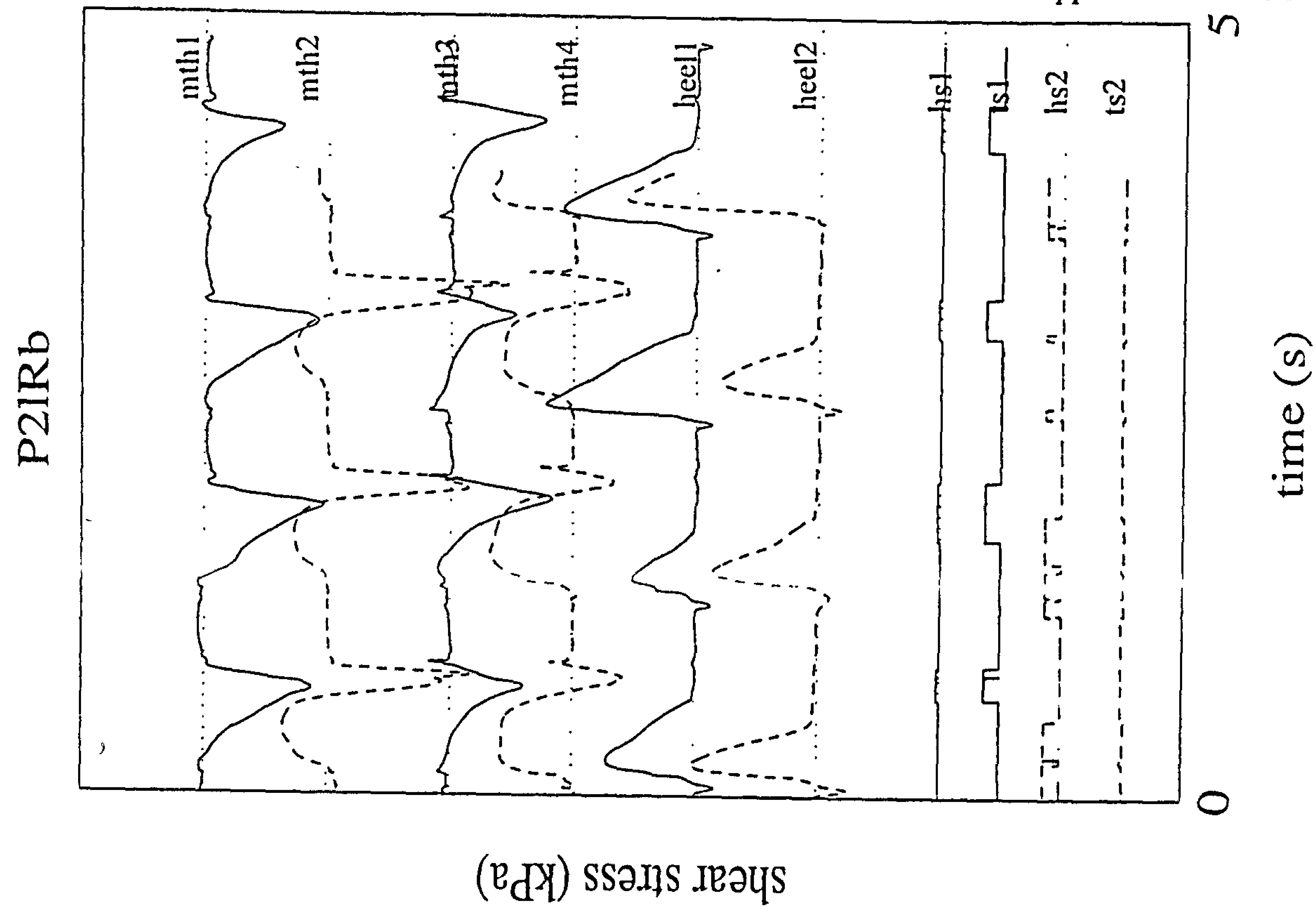
The shear records of repeat tests are referred to by the addition of the letter *R* to the end of the code described above, eg. *P7lRs-R*.

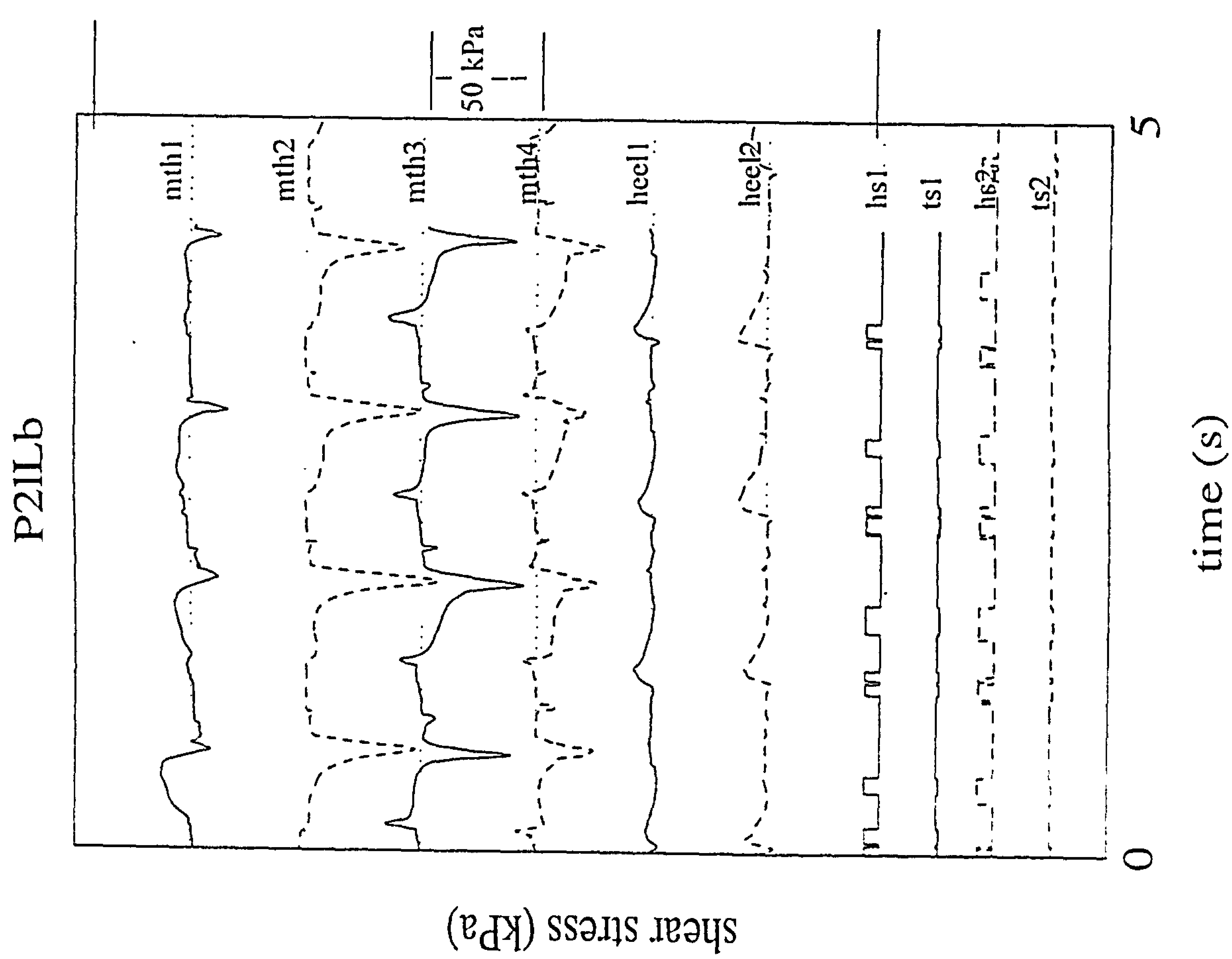

P11Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

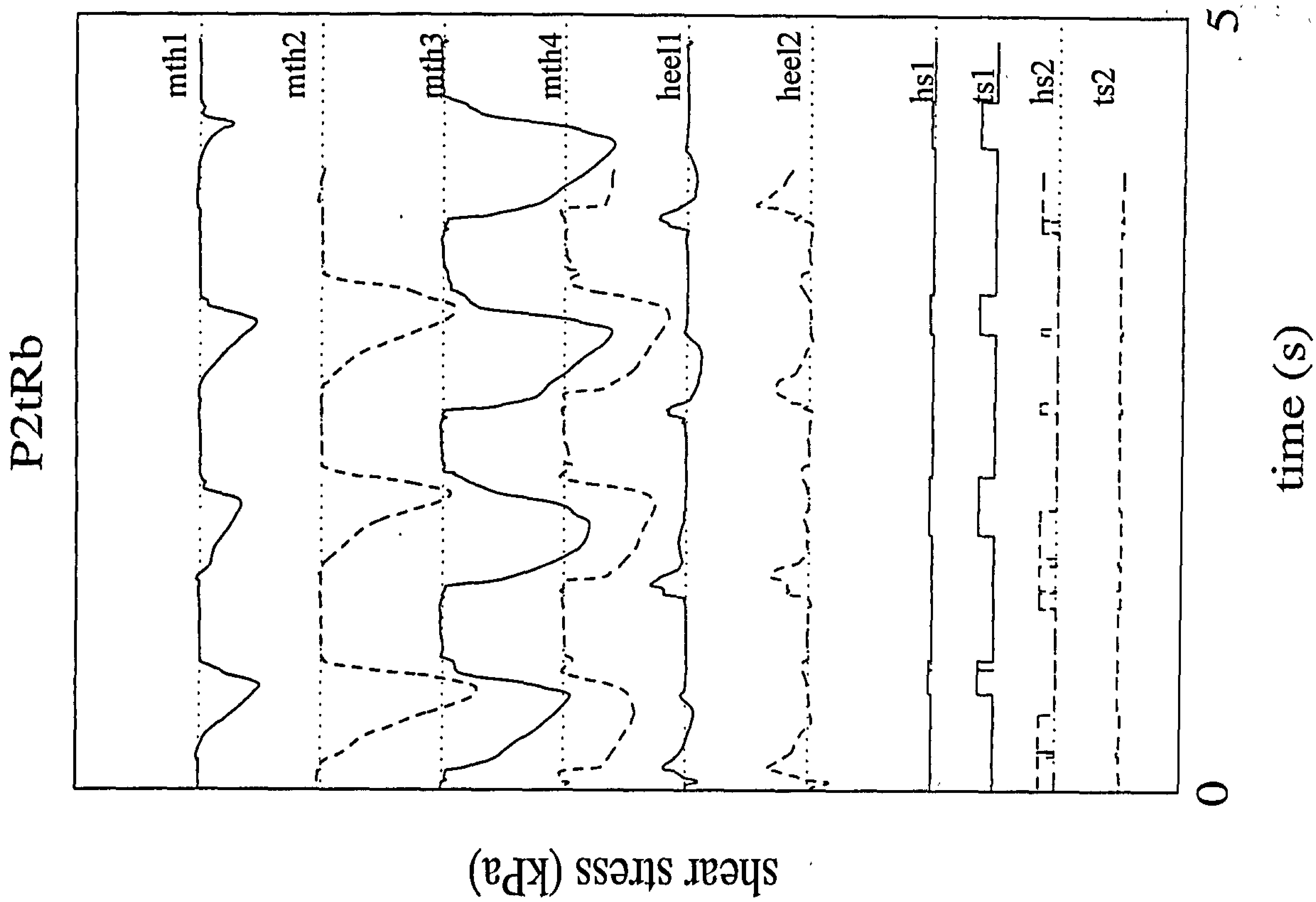

P11Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

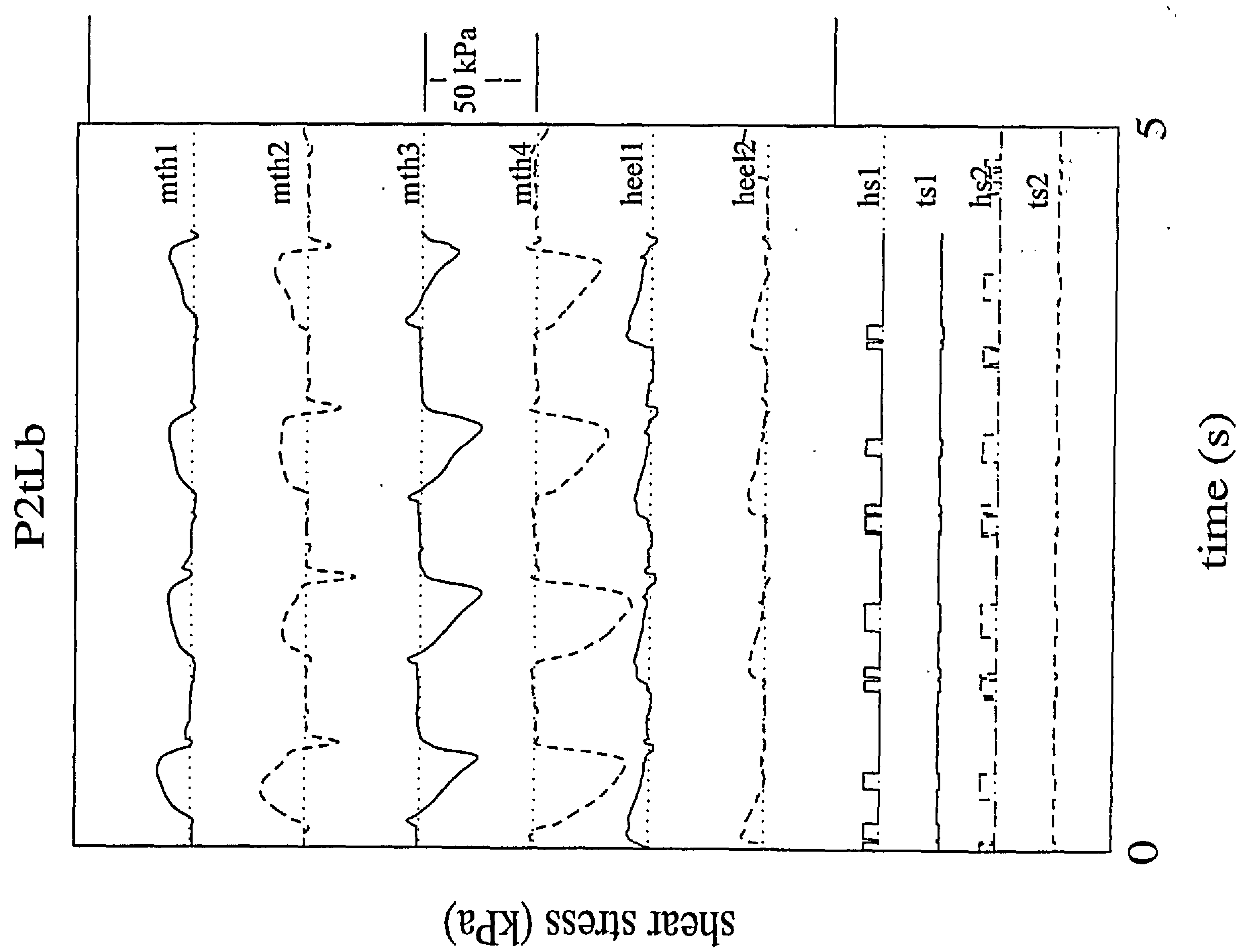

P1tLb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
time (s)
0
5

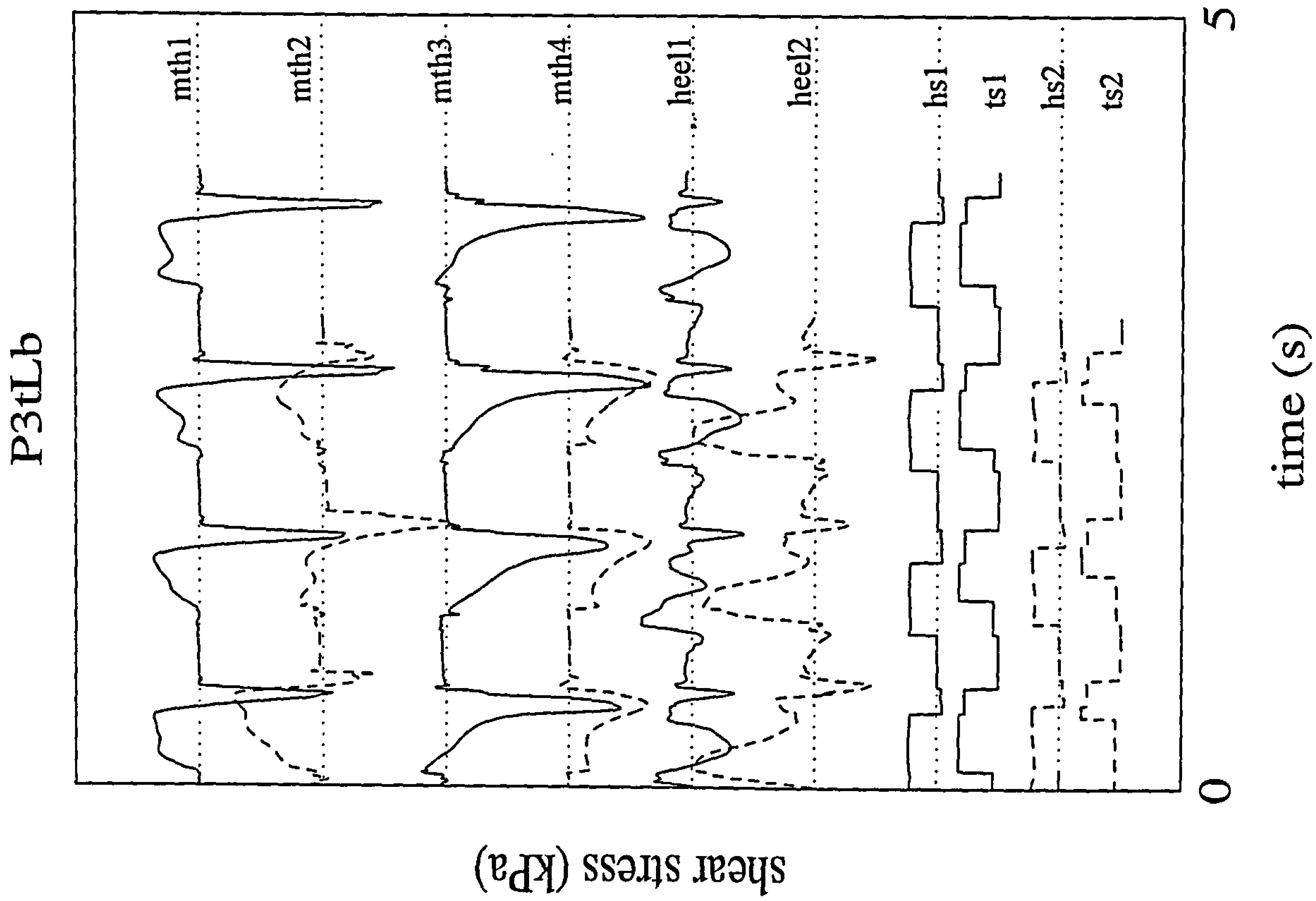

P1tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

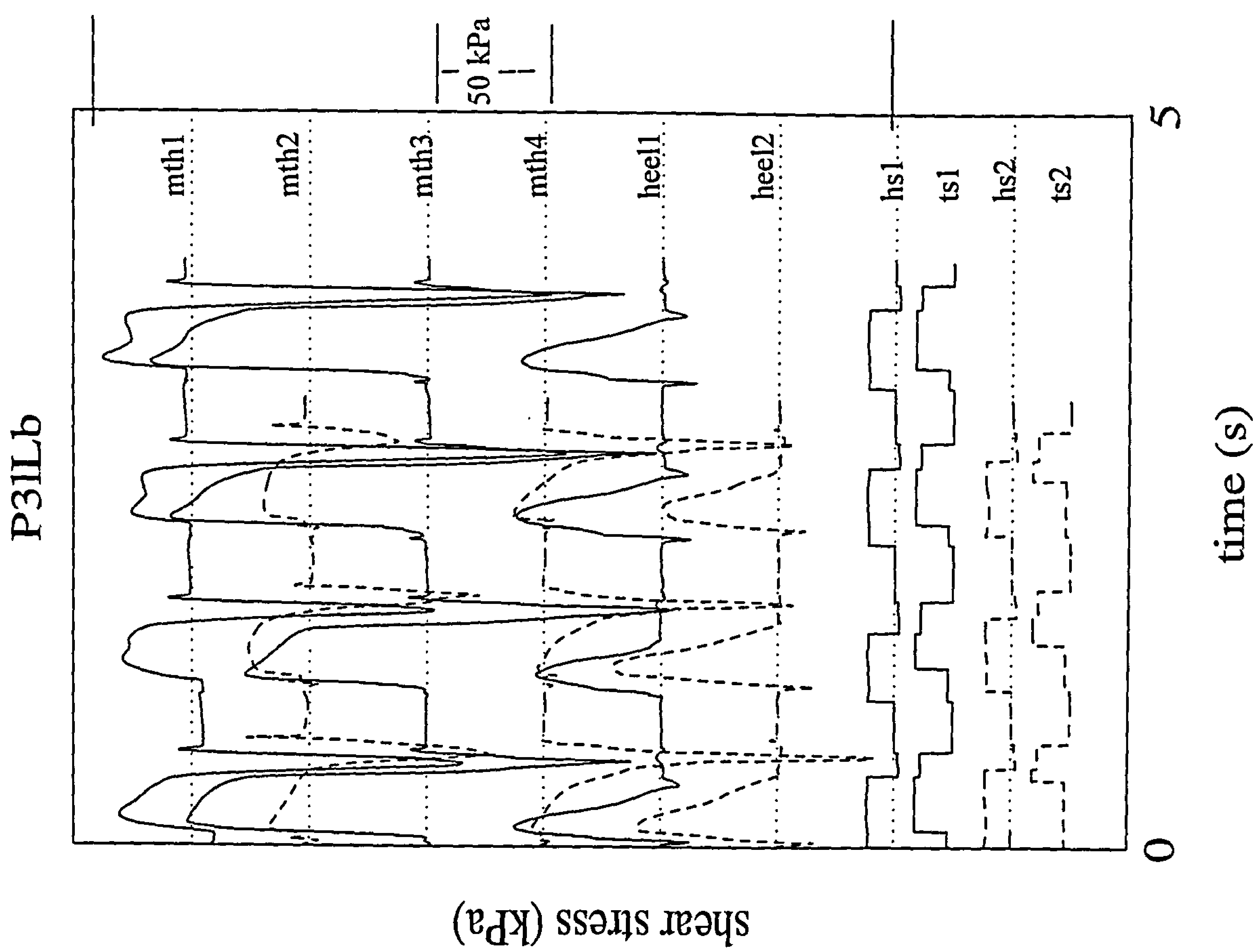

P21Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

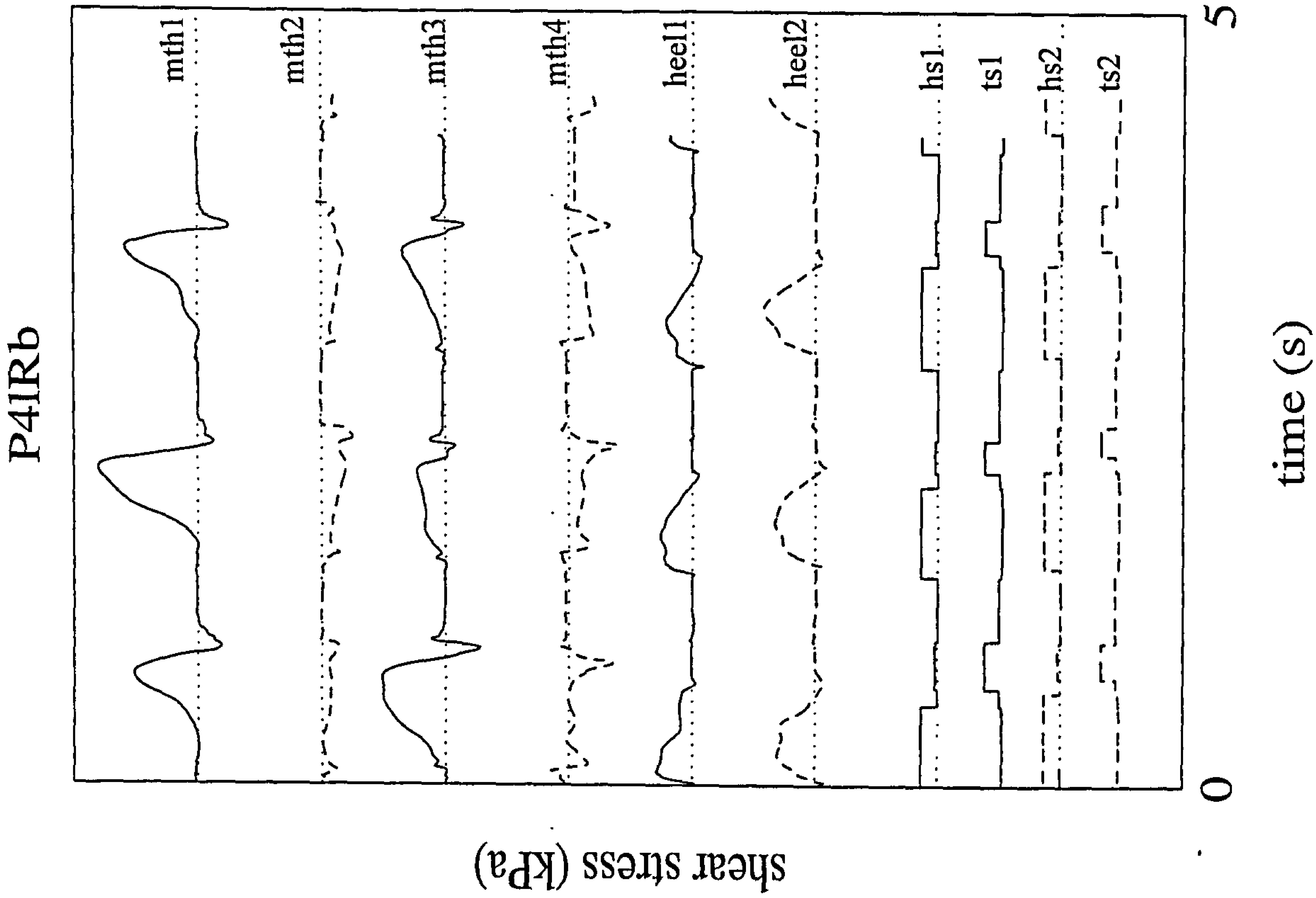

P21Lb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
time (s)
0
5

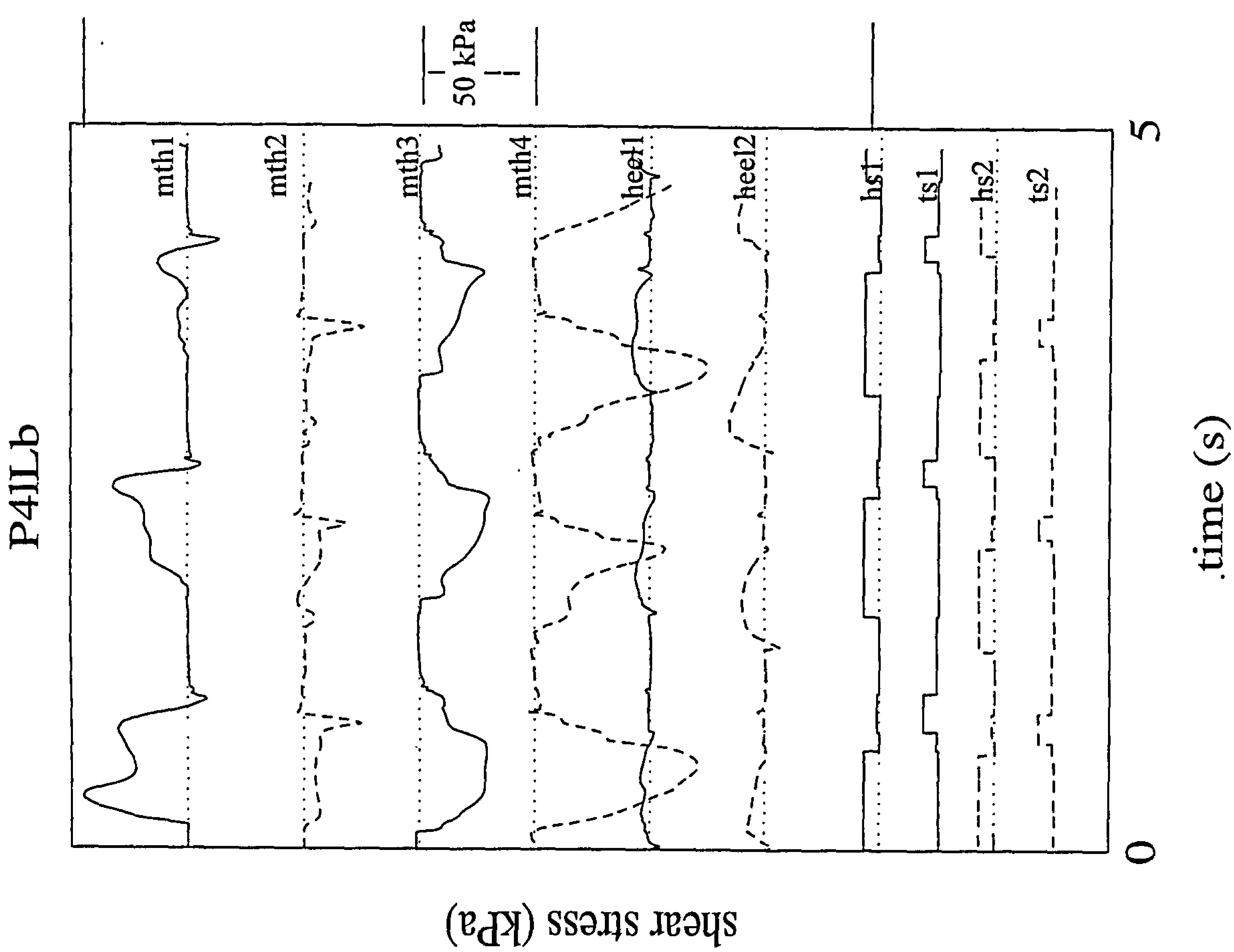

P2tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

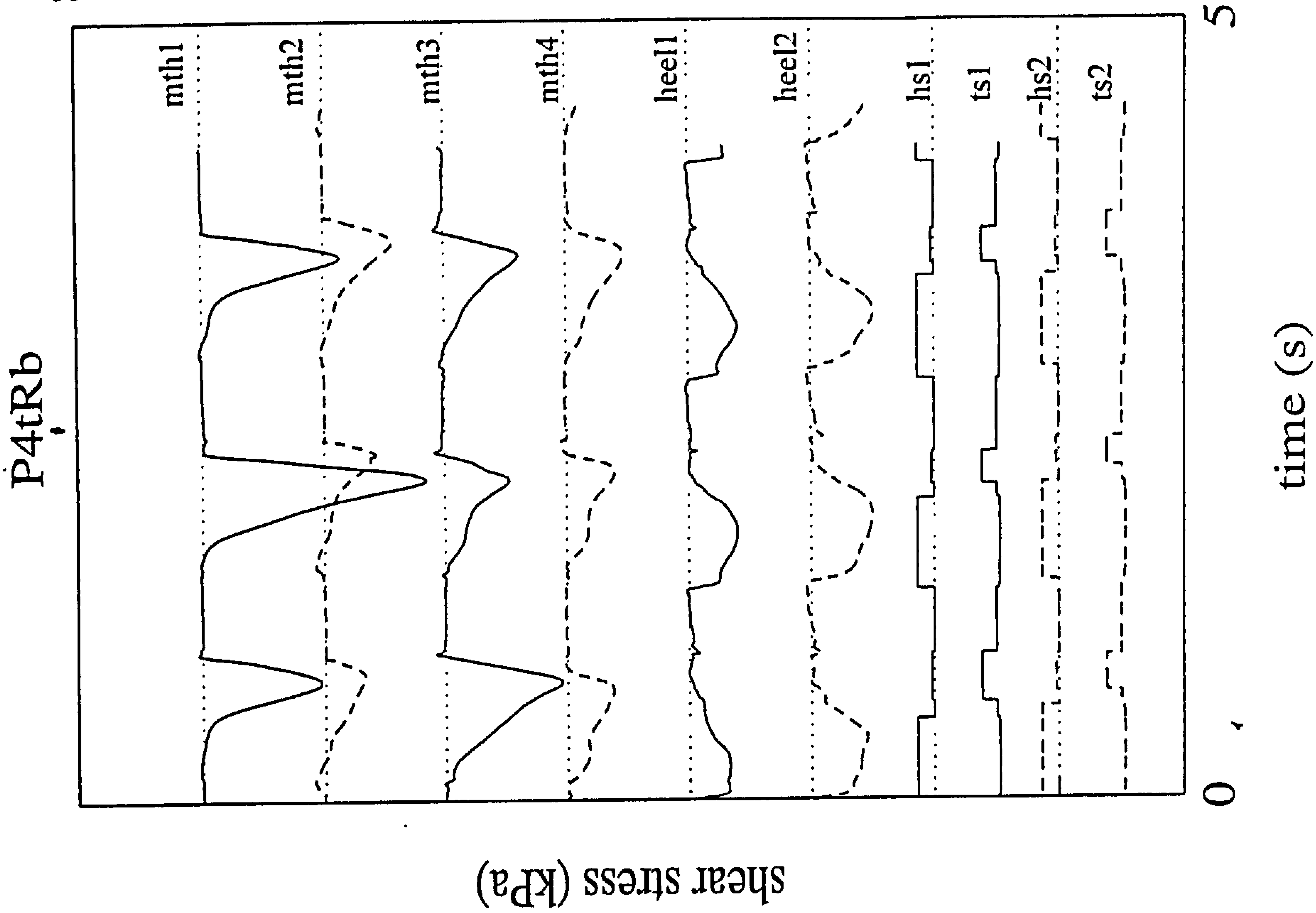

P2tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

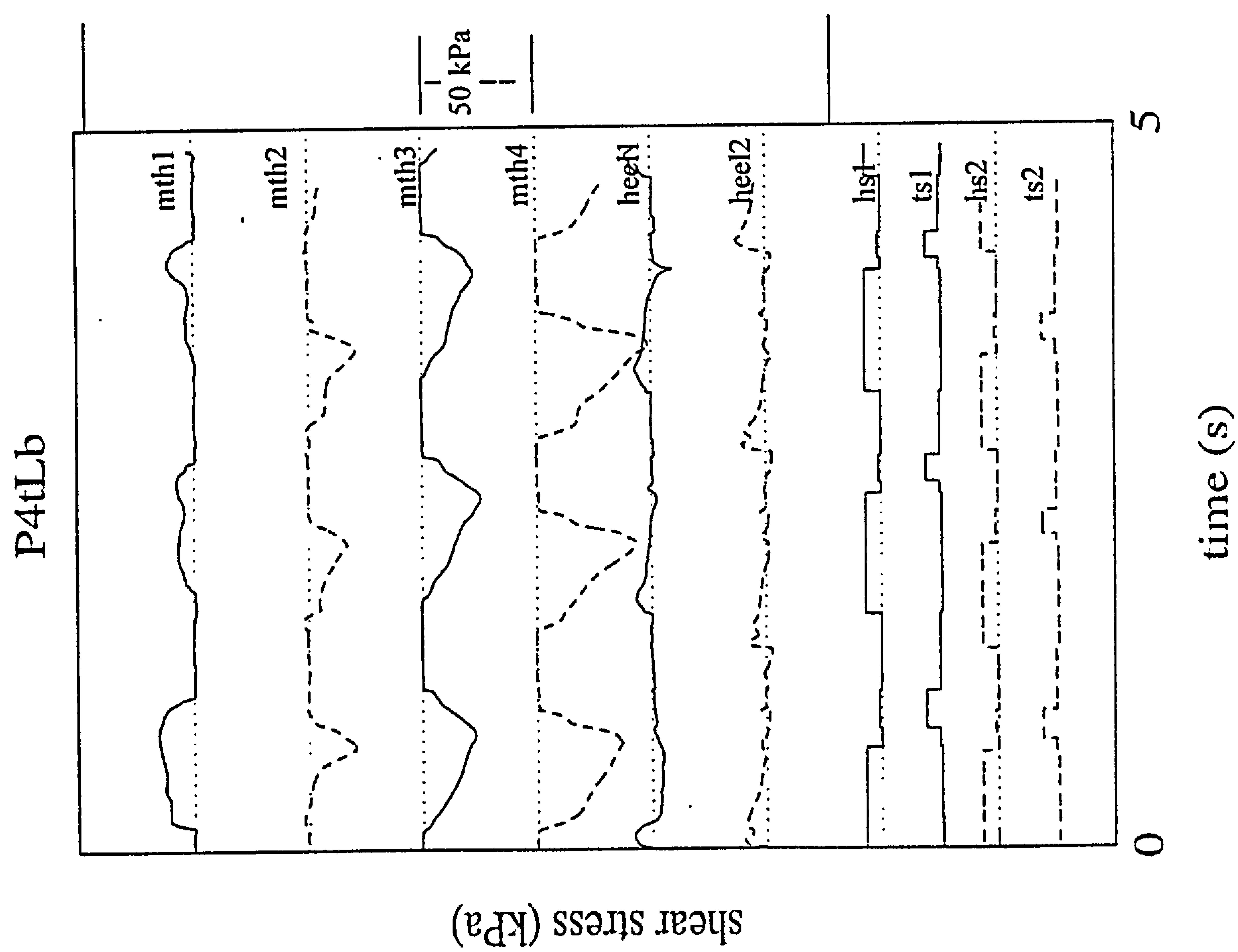

P3tLb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

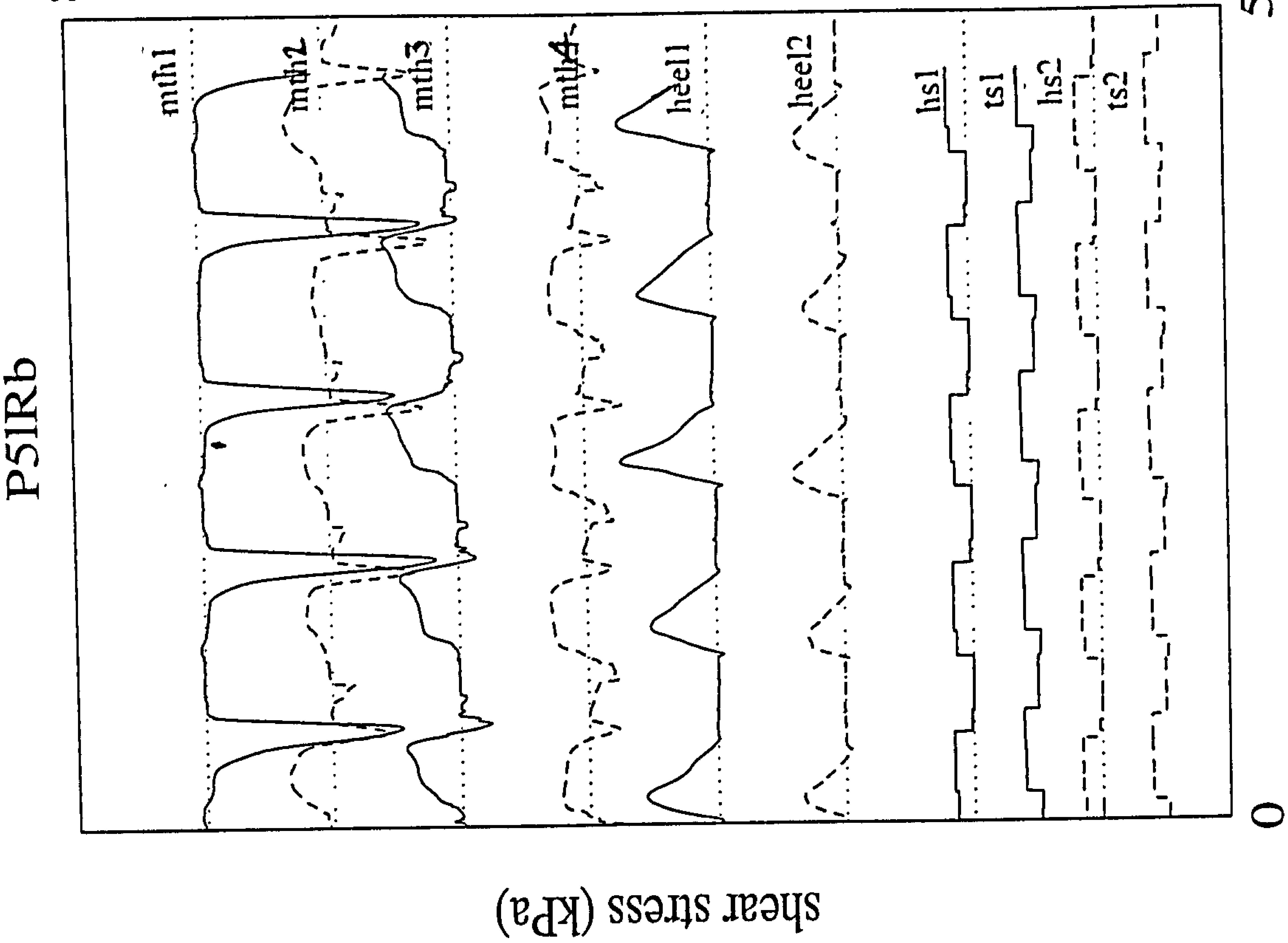

P31Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

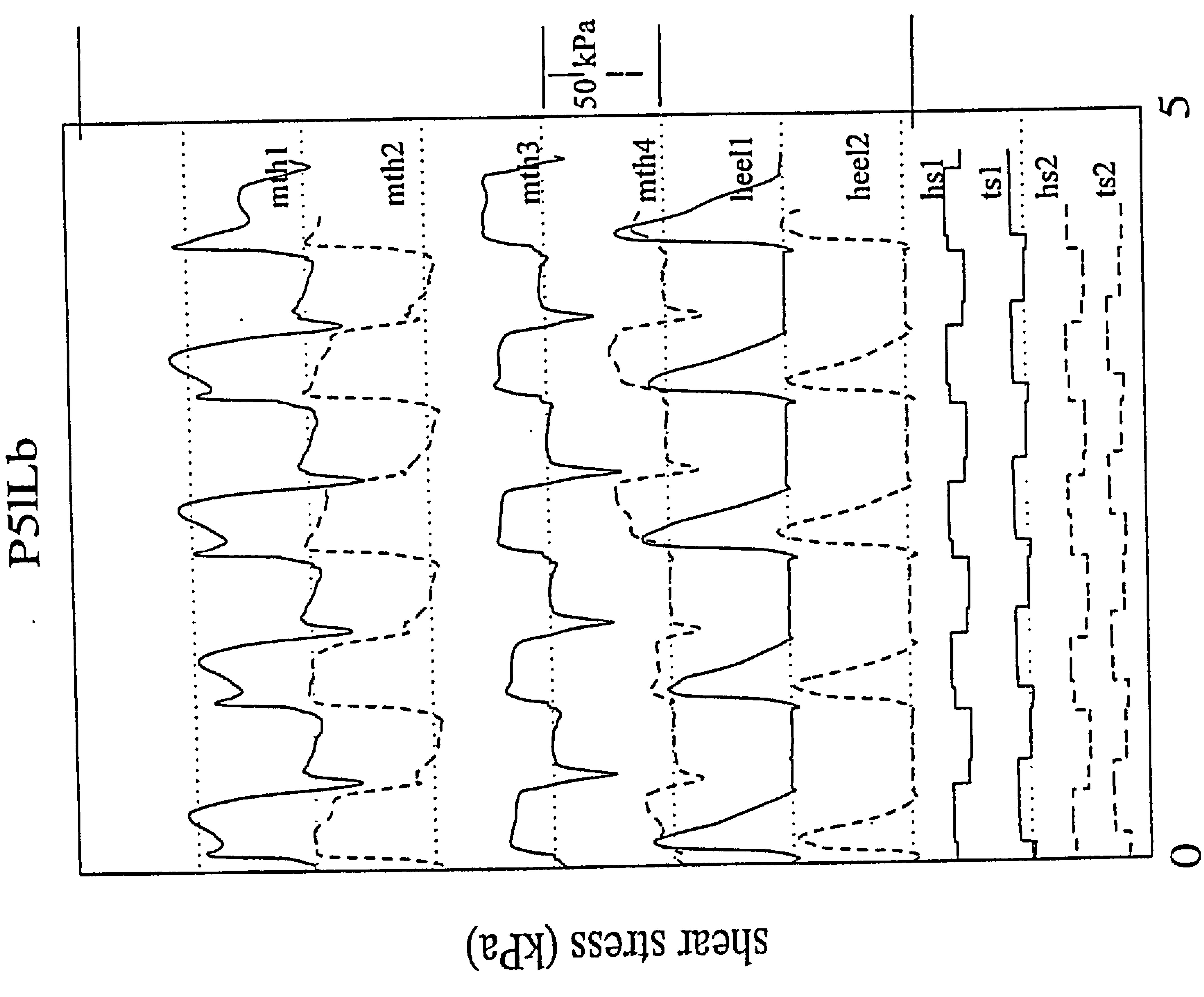

P41Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

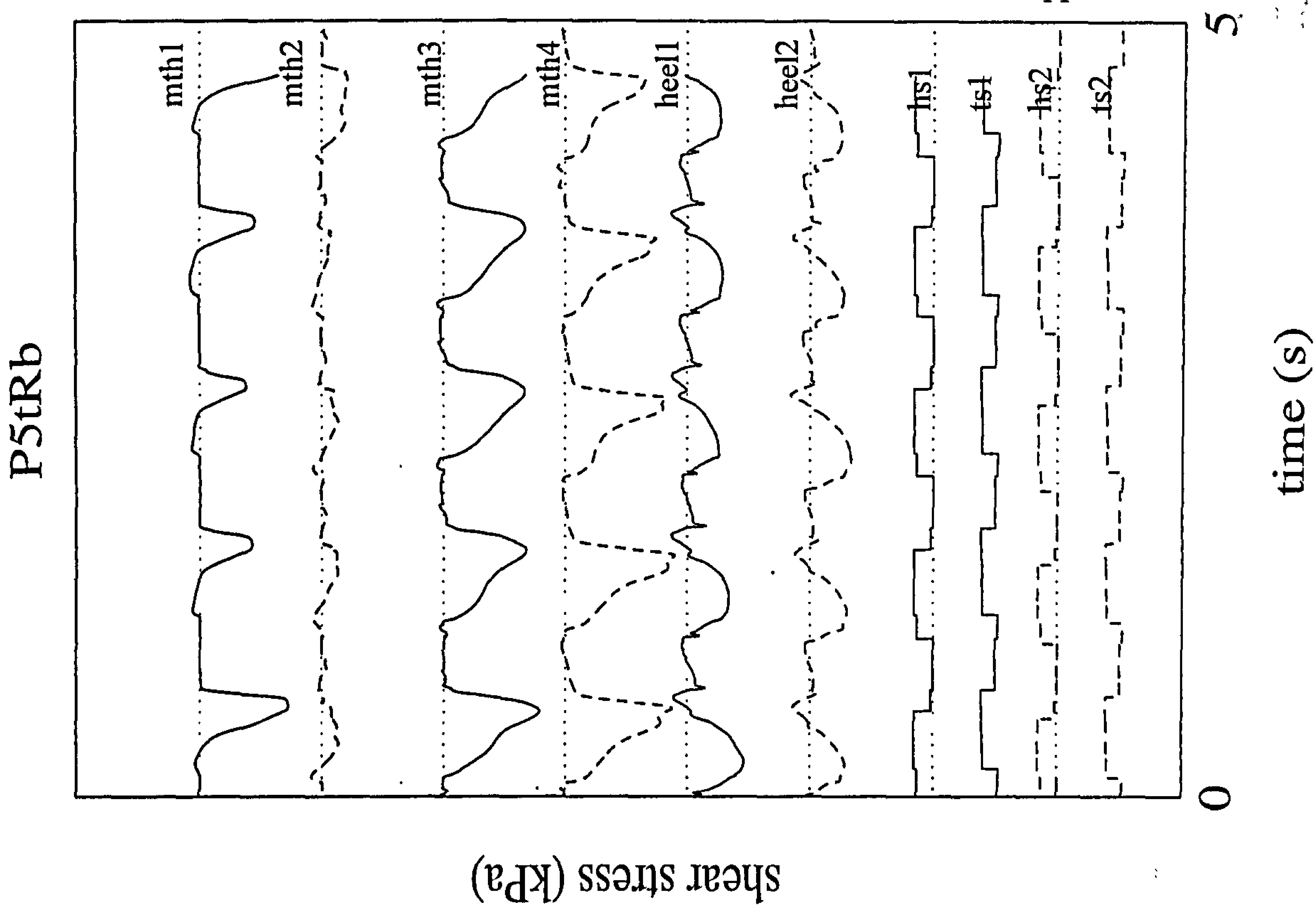

P41Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

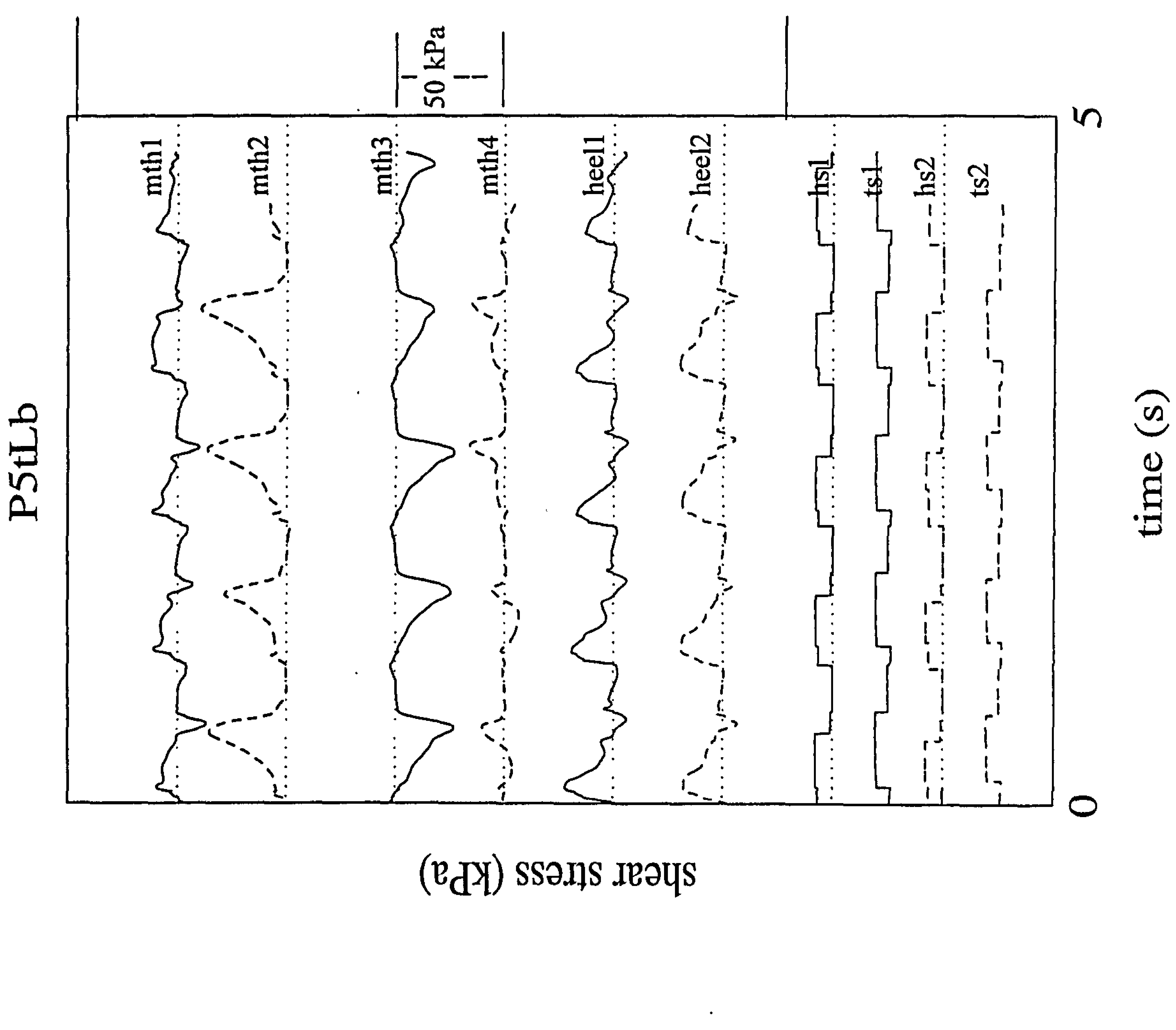

P4tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

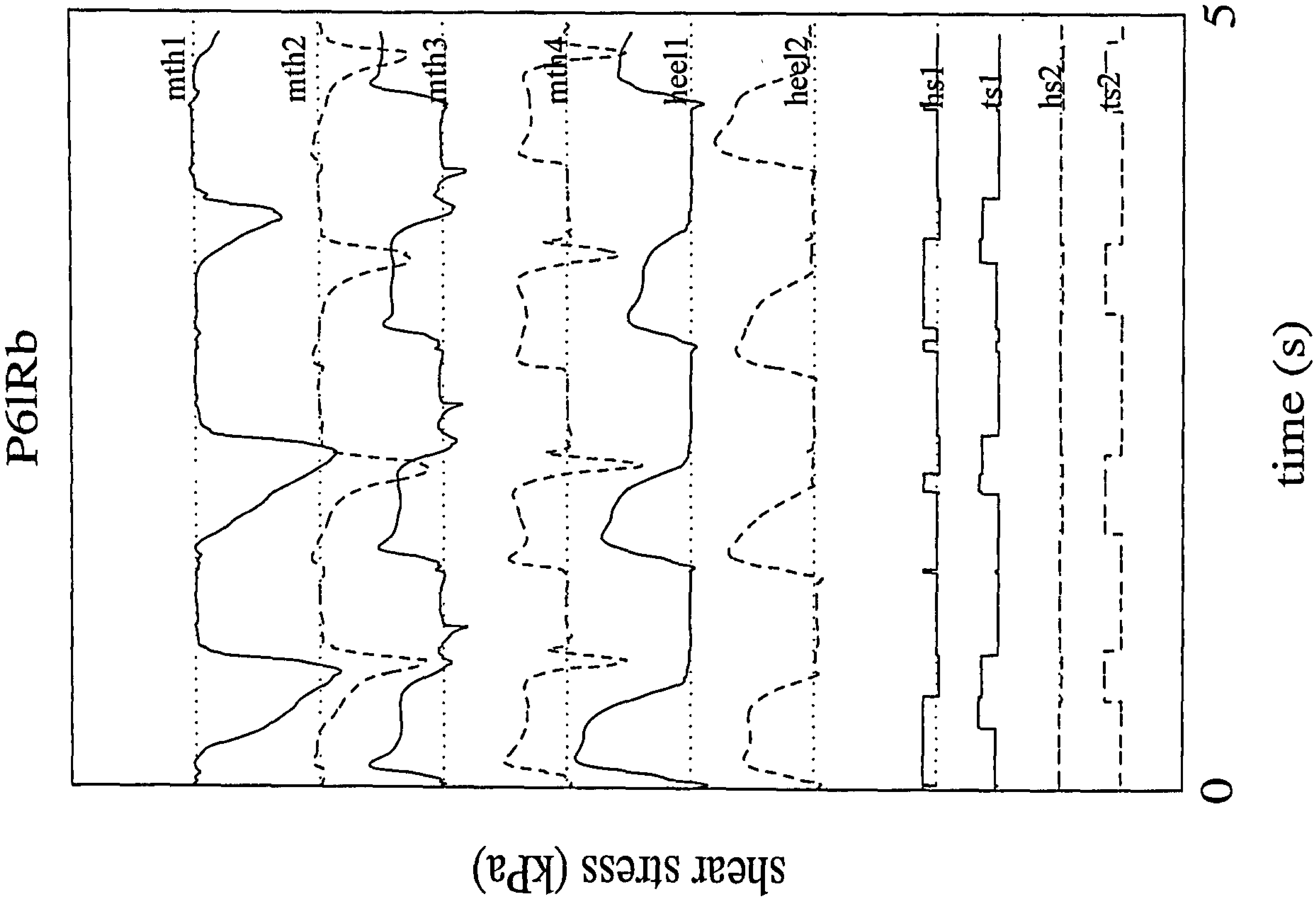

P4tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

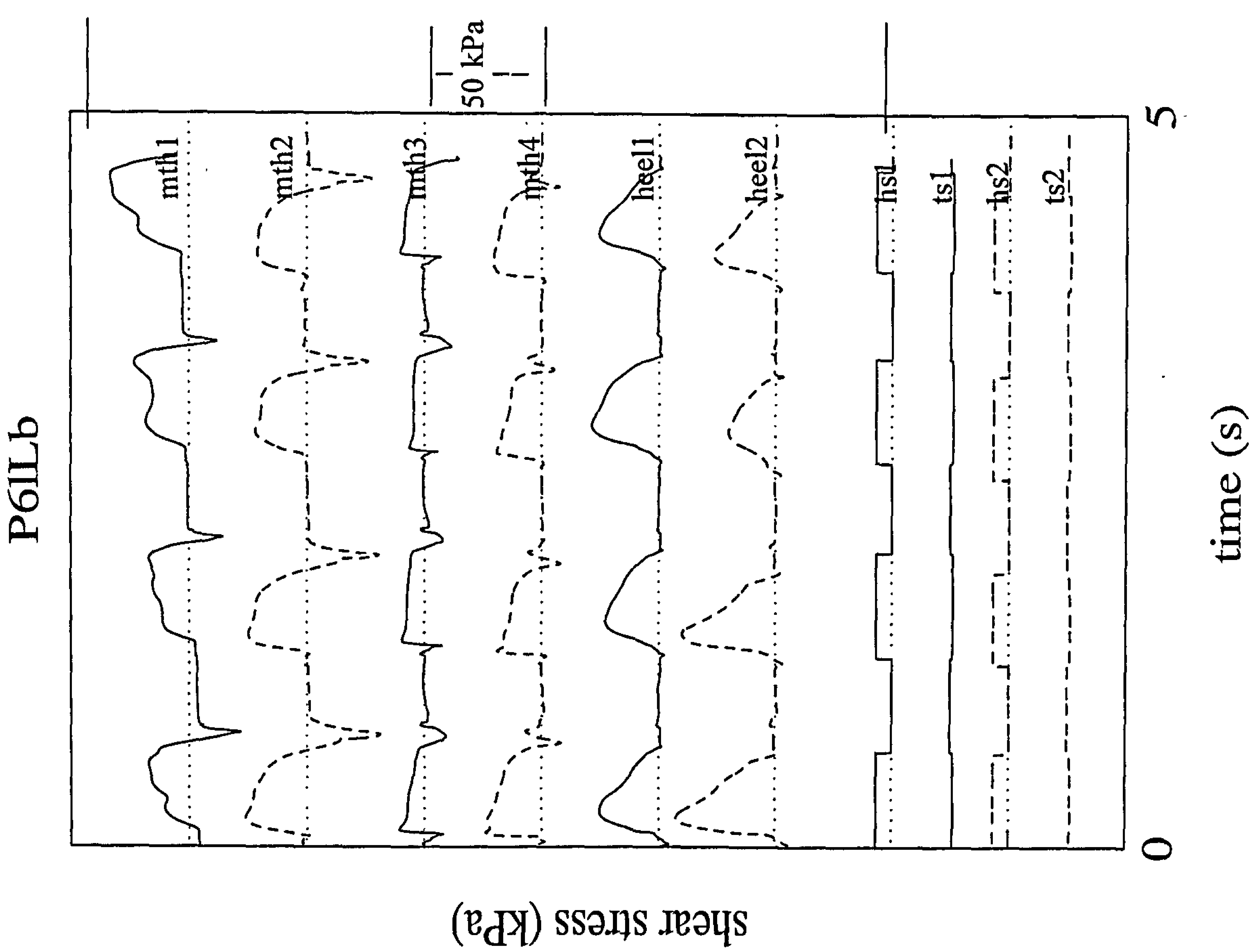

P51Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

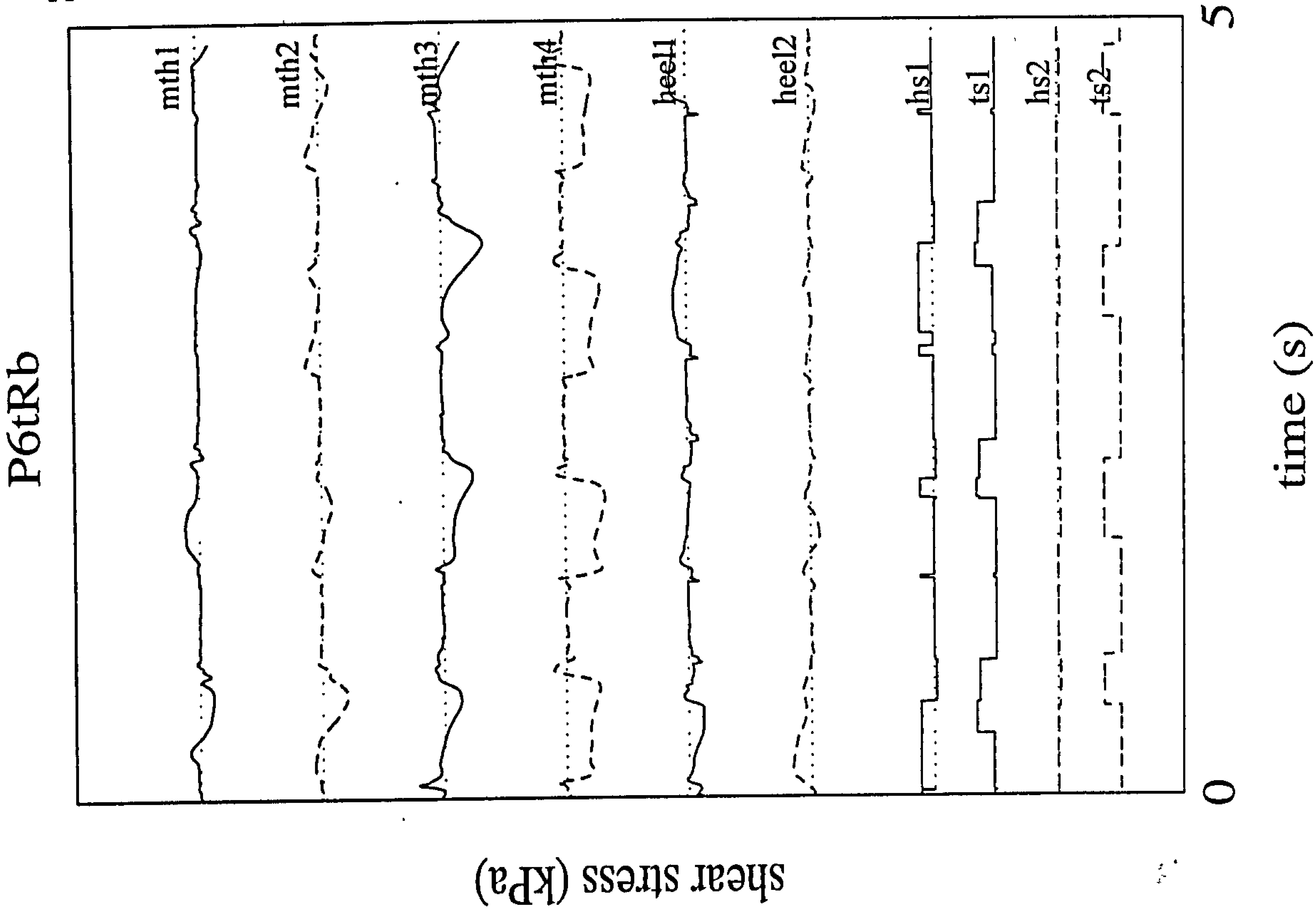

P51Lb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
0
5
time (s)
shear stress (kPa)

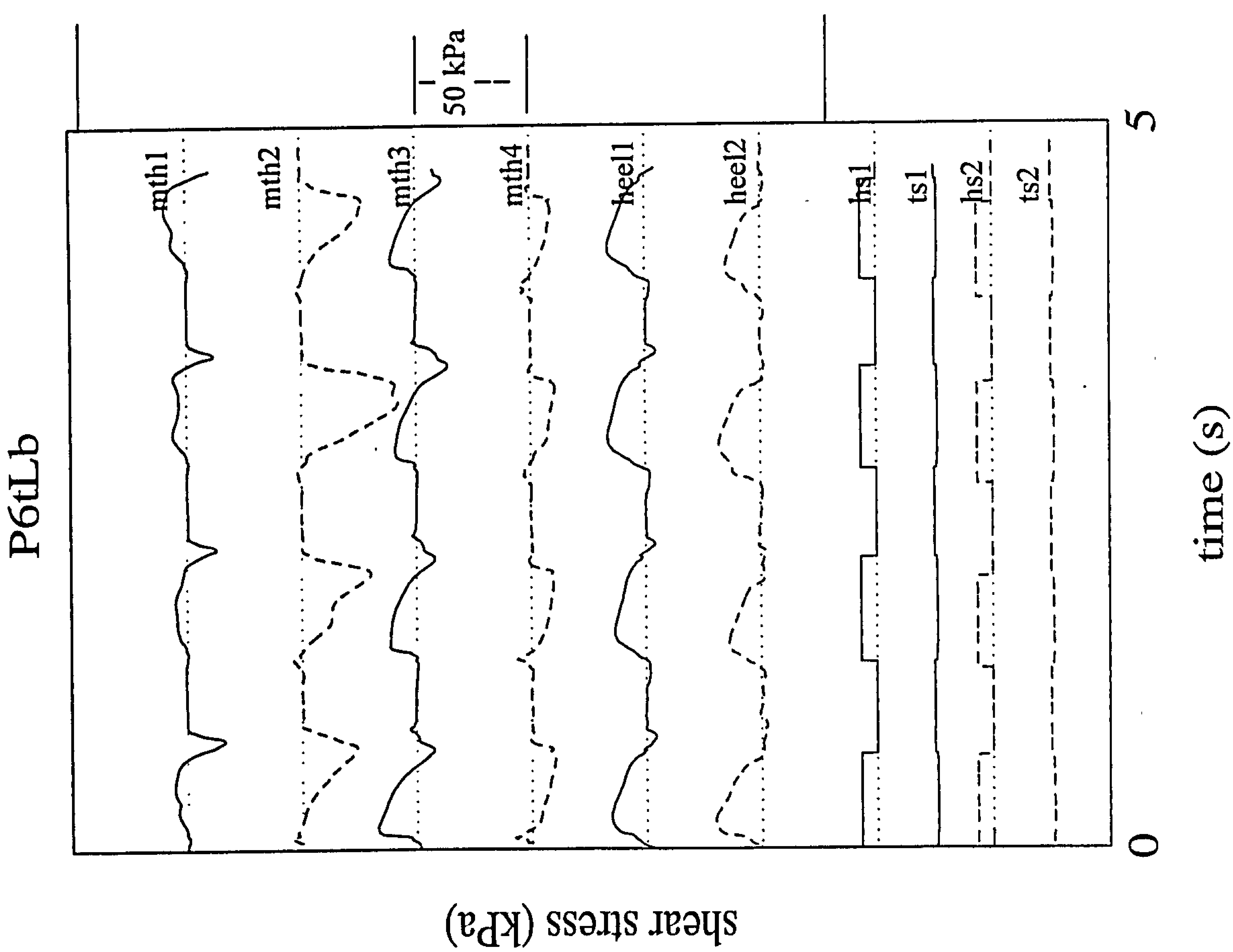

P5tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)

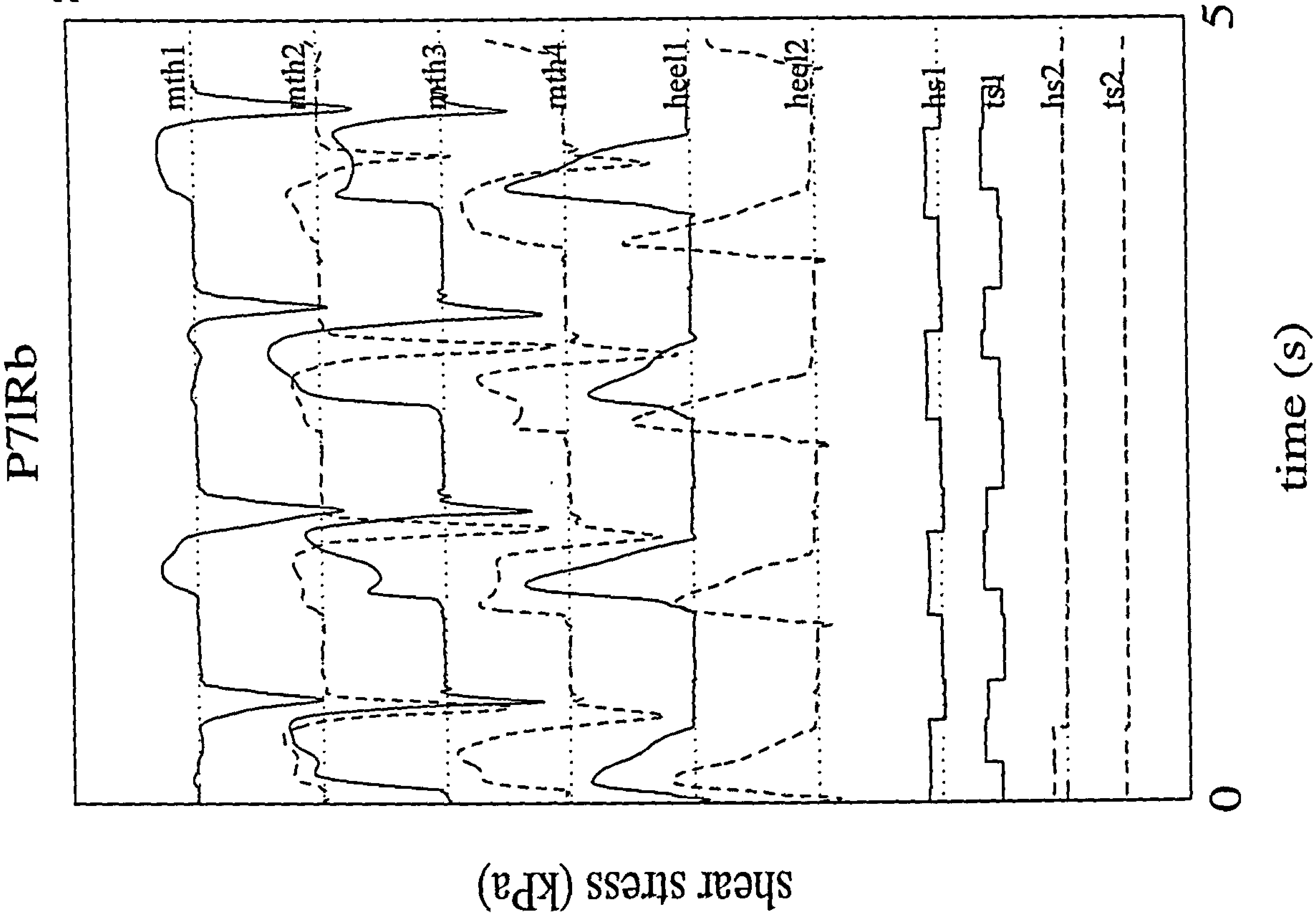

P5tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)

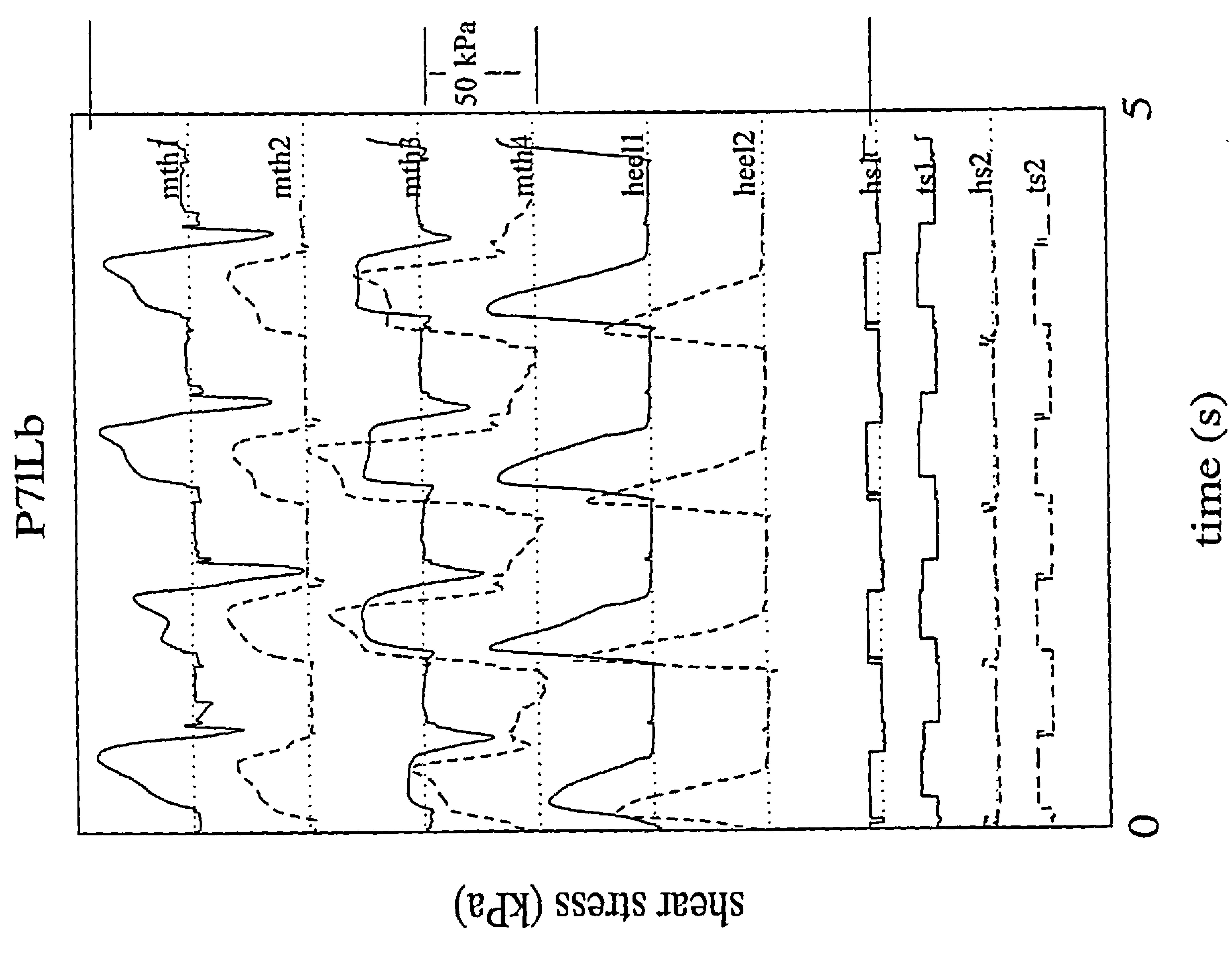

P61Rb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)

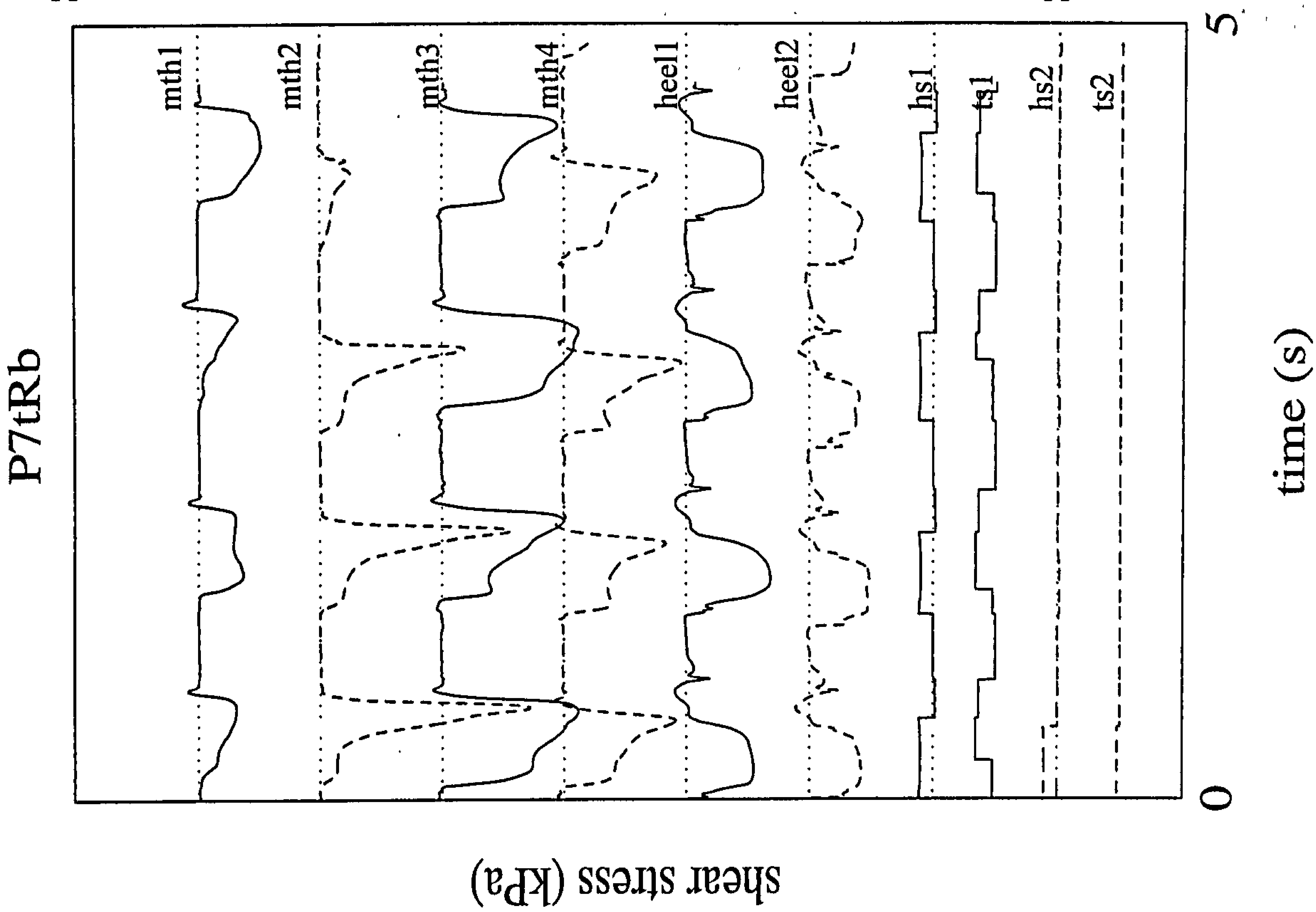

P61Lb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)

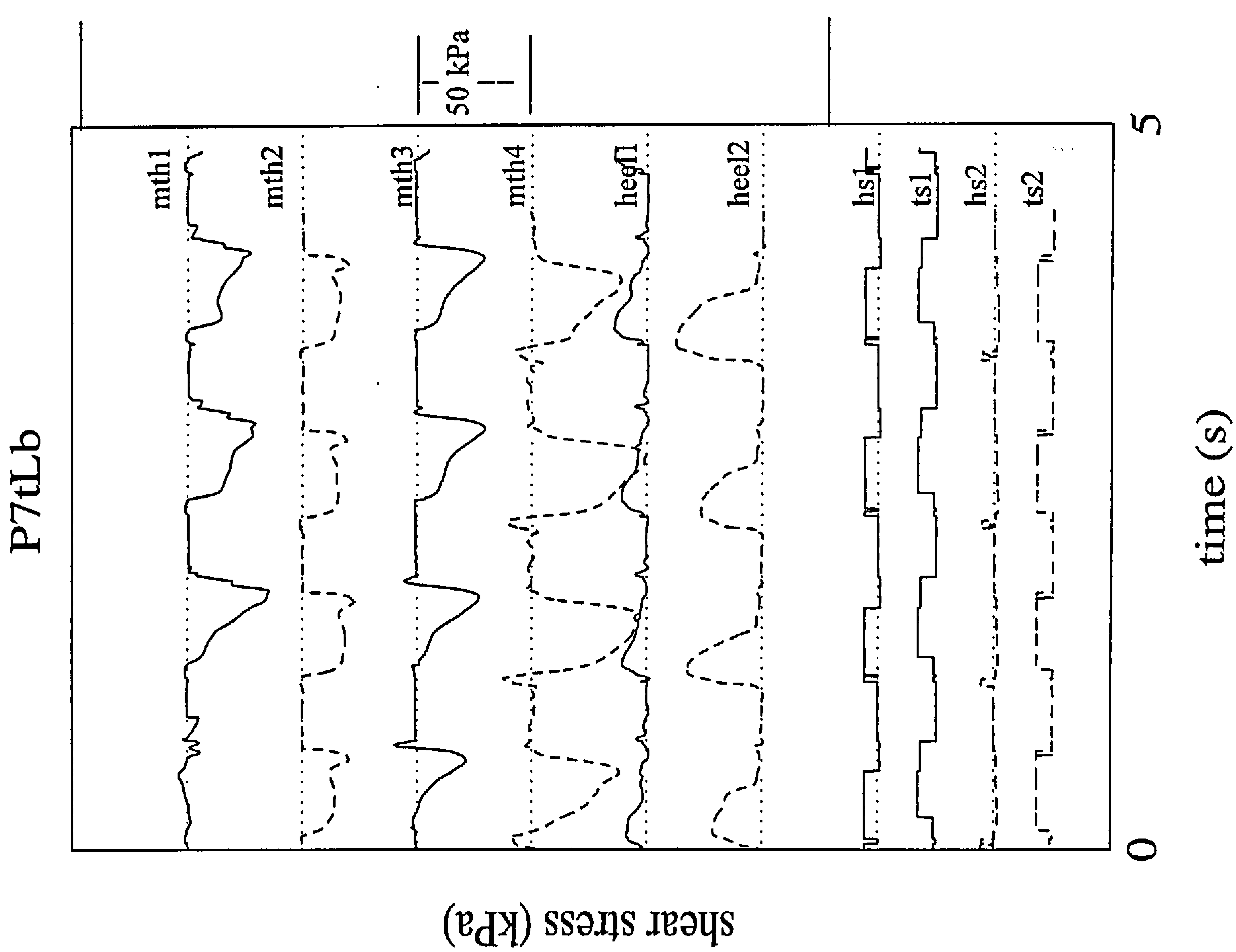

P6tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

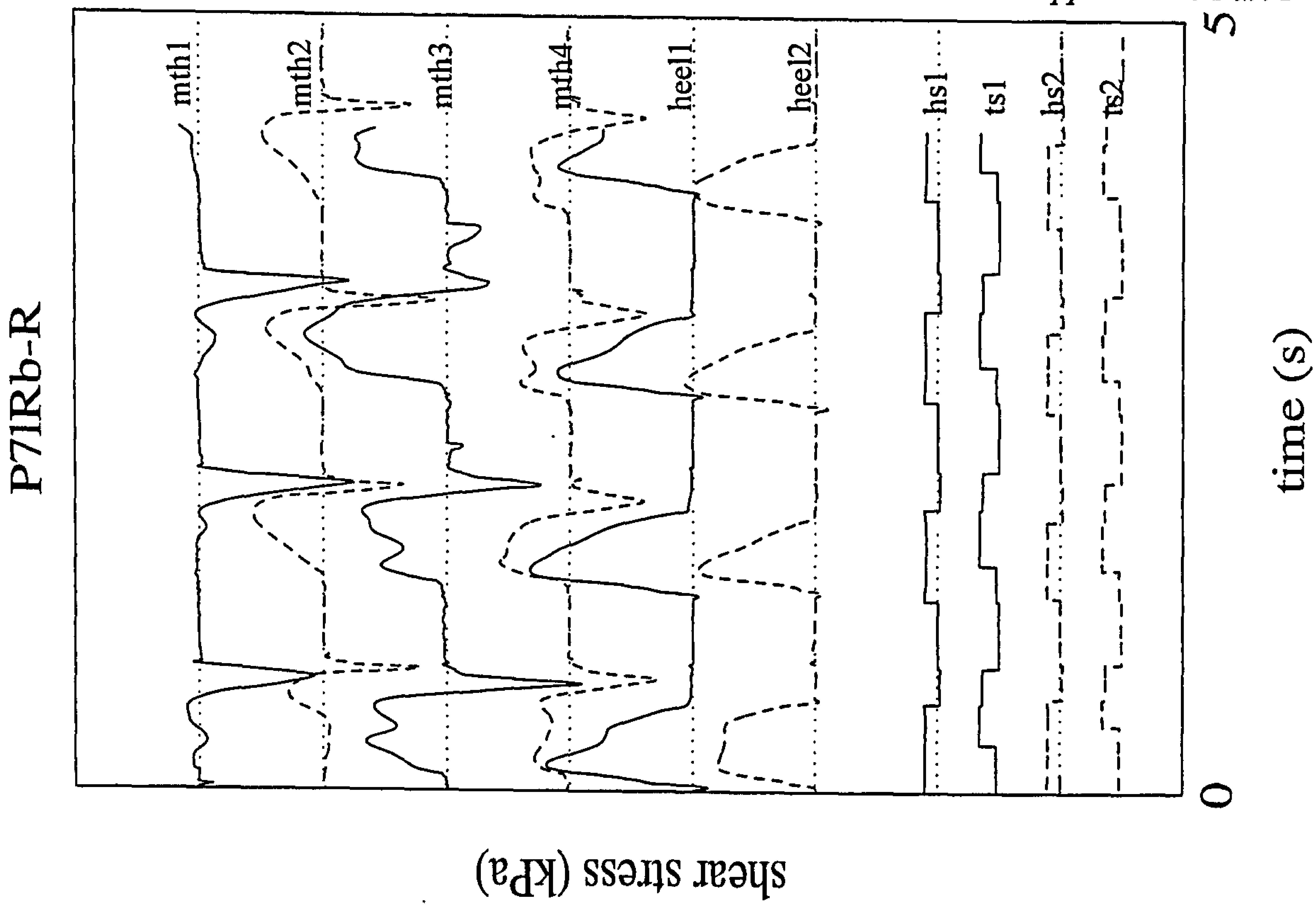

P6tLb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
time (s)
0
5

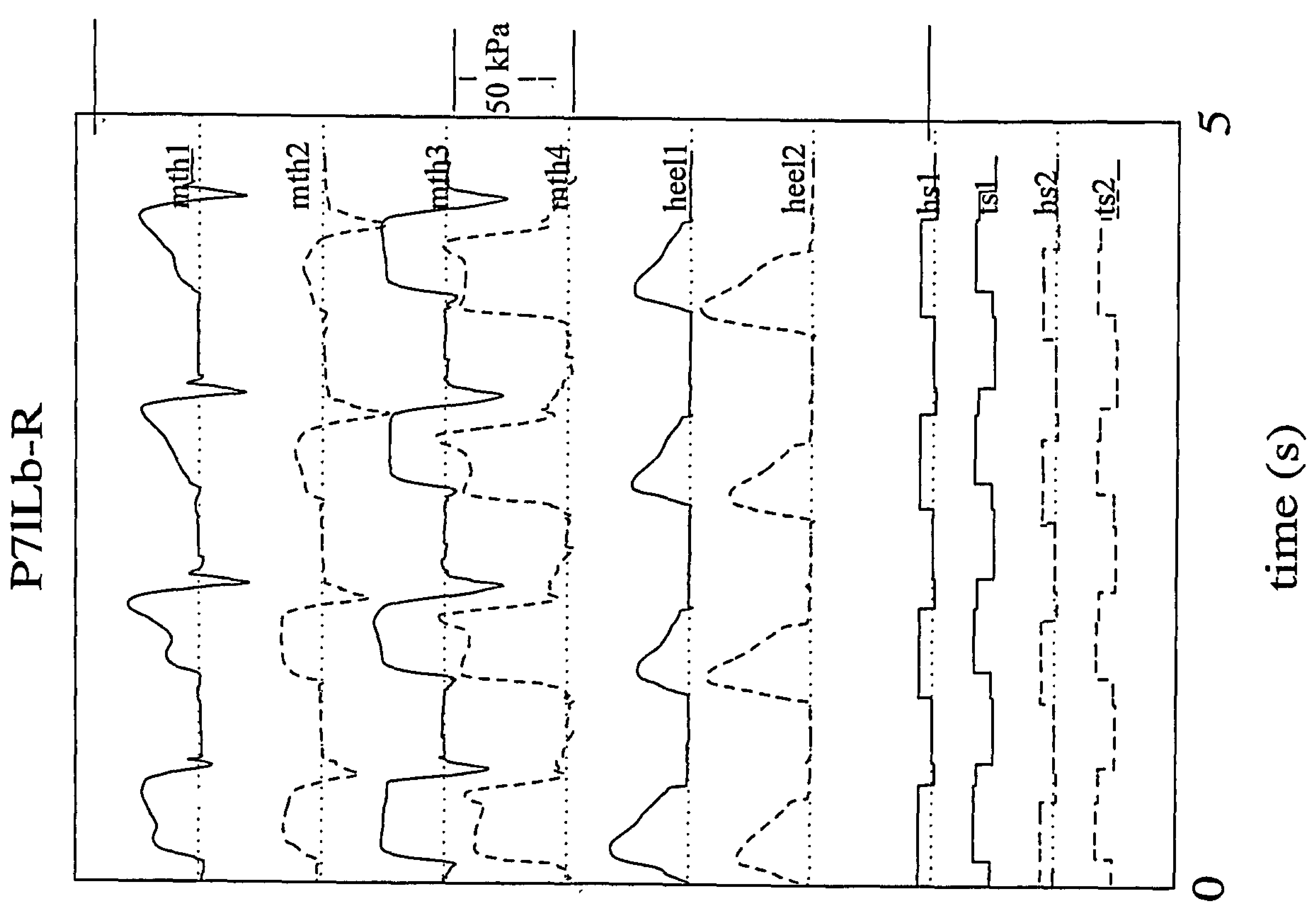

P7IRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

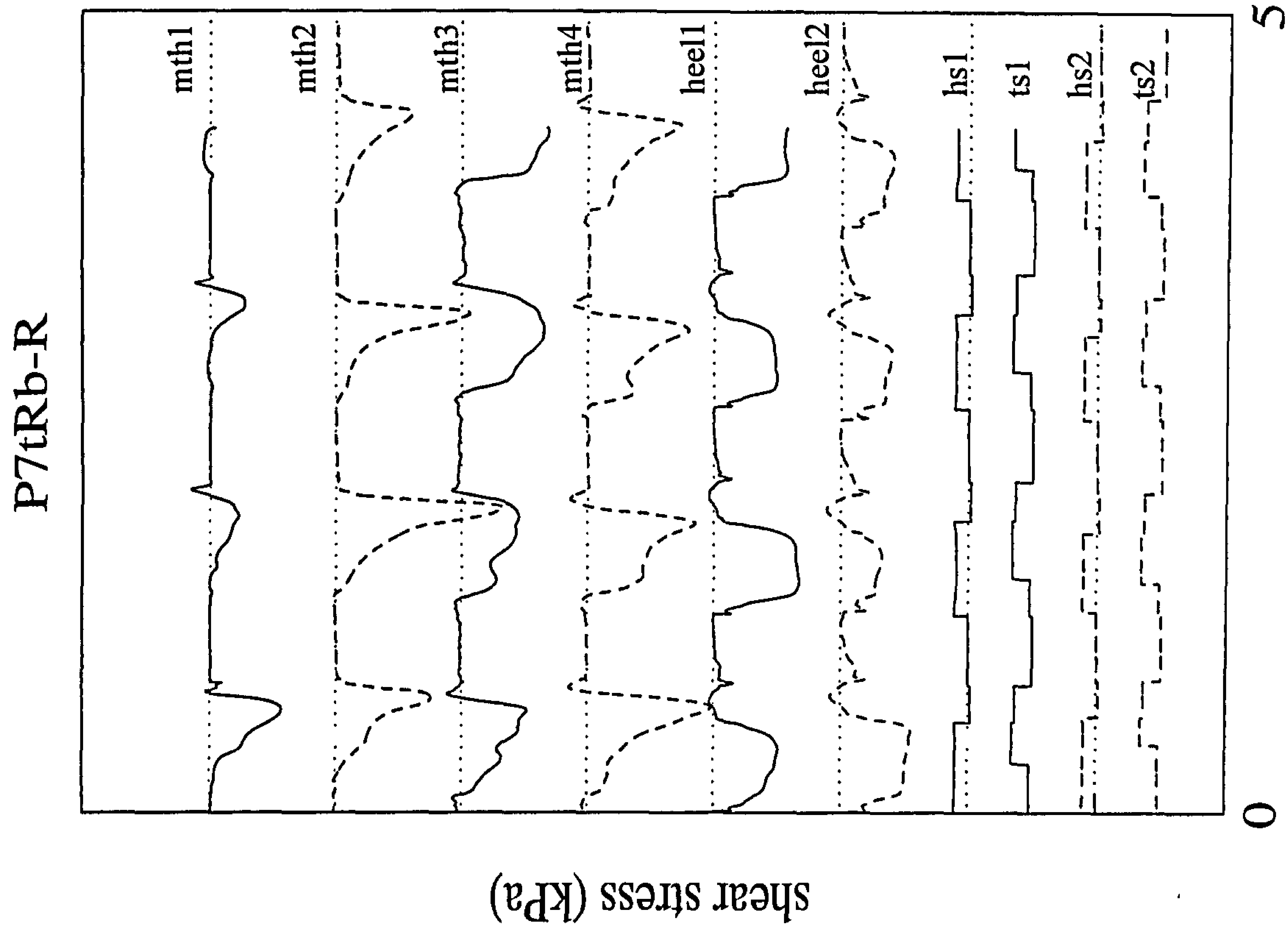

P7ILb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

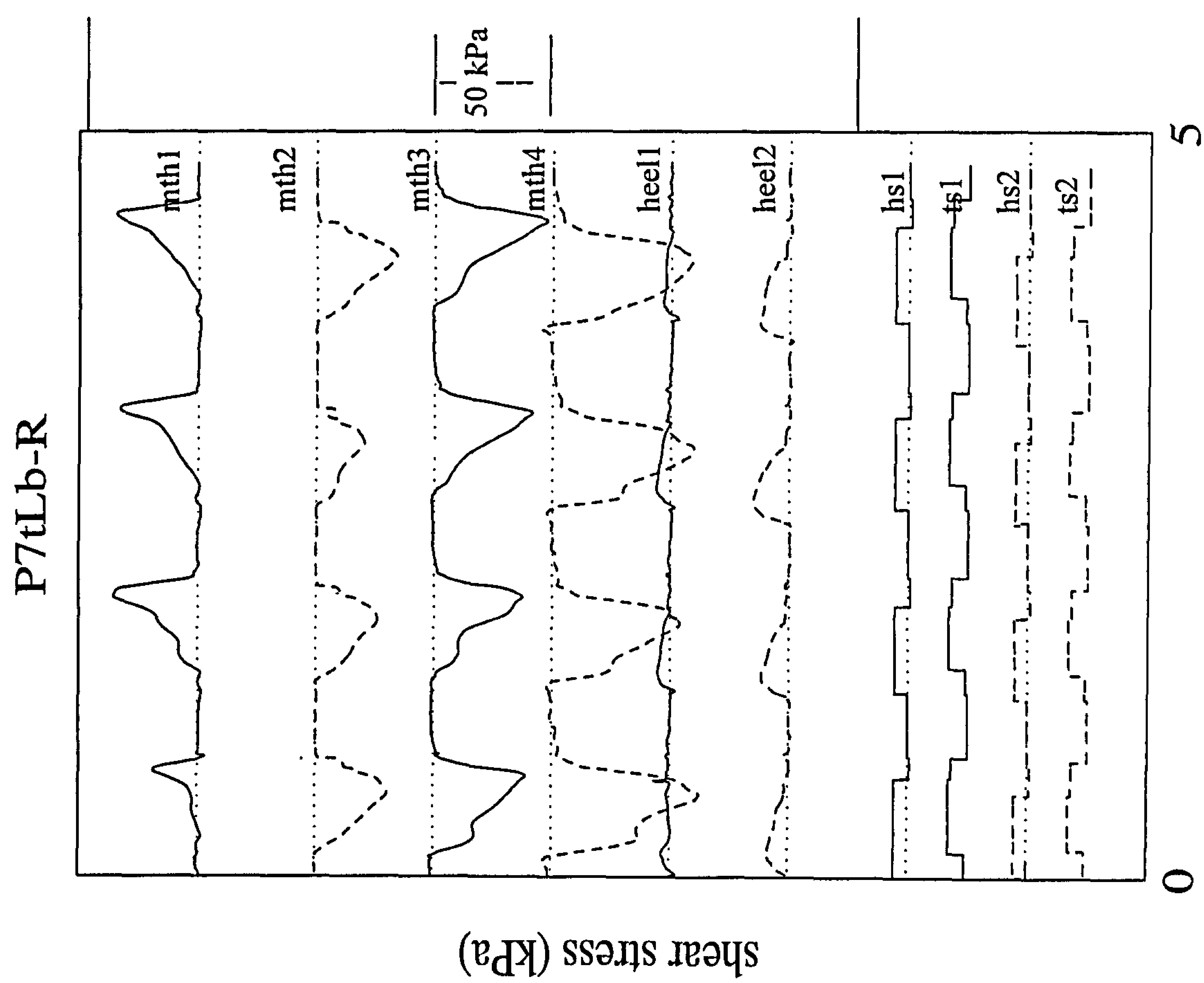

P7tRb
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)

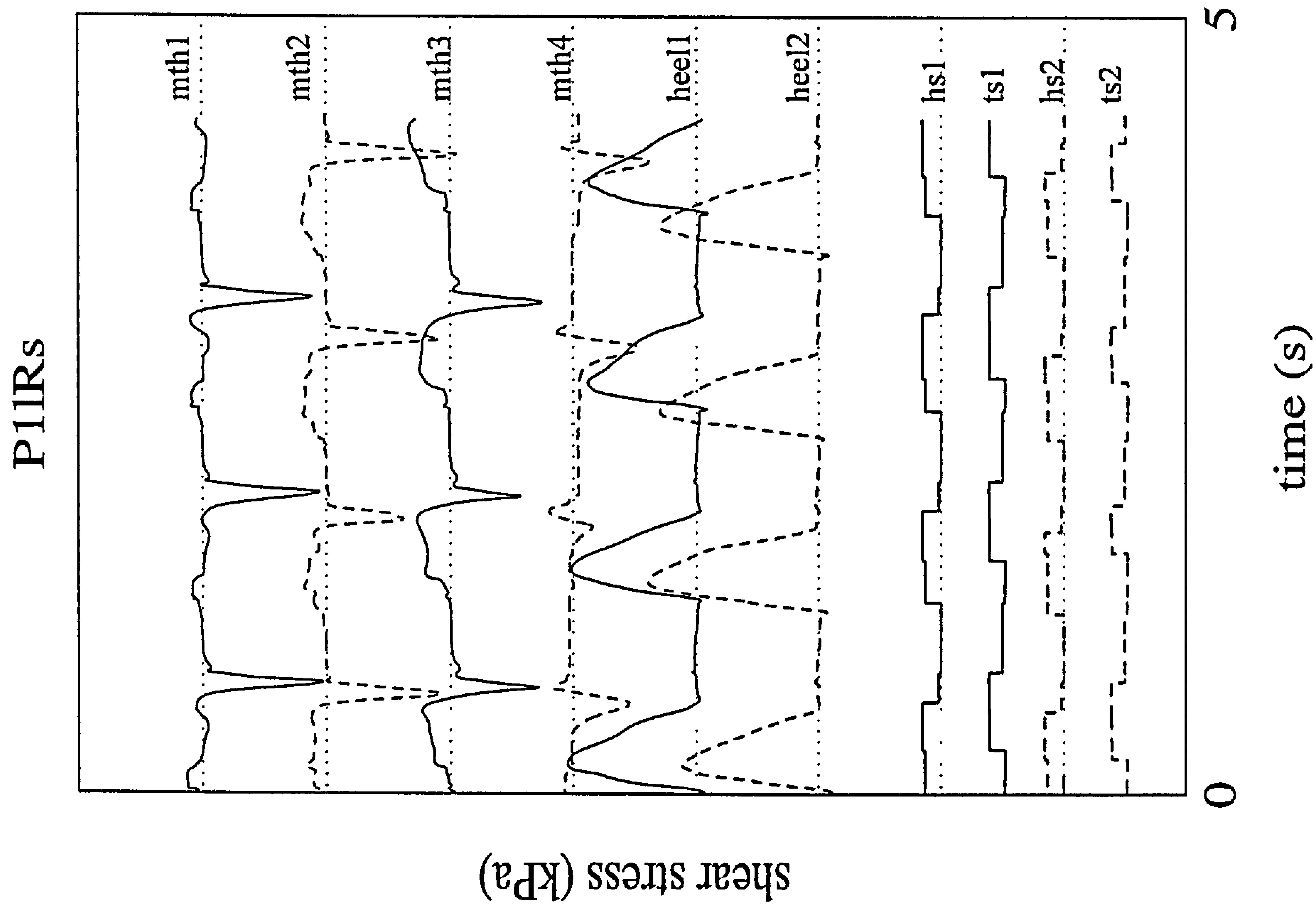

P7tLb
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)

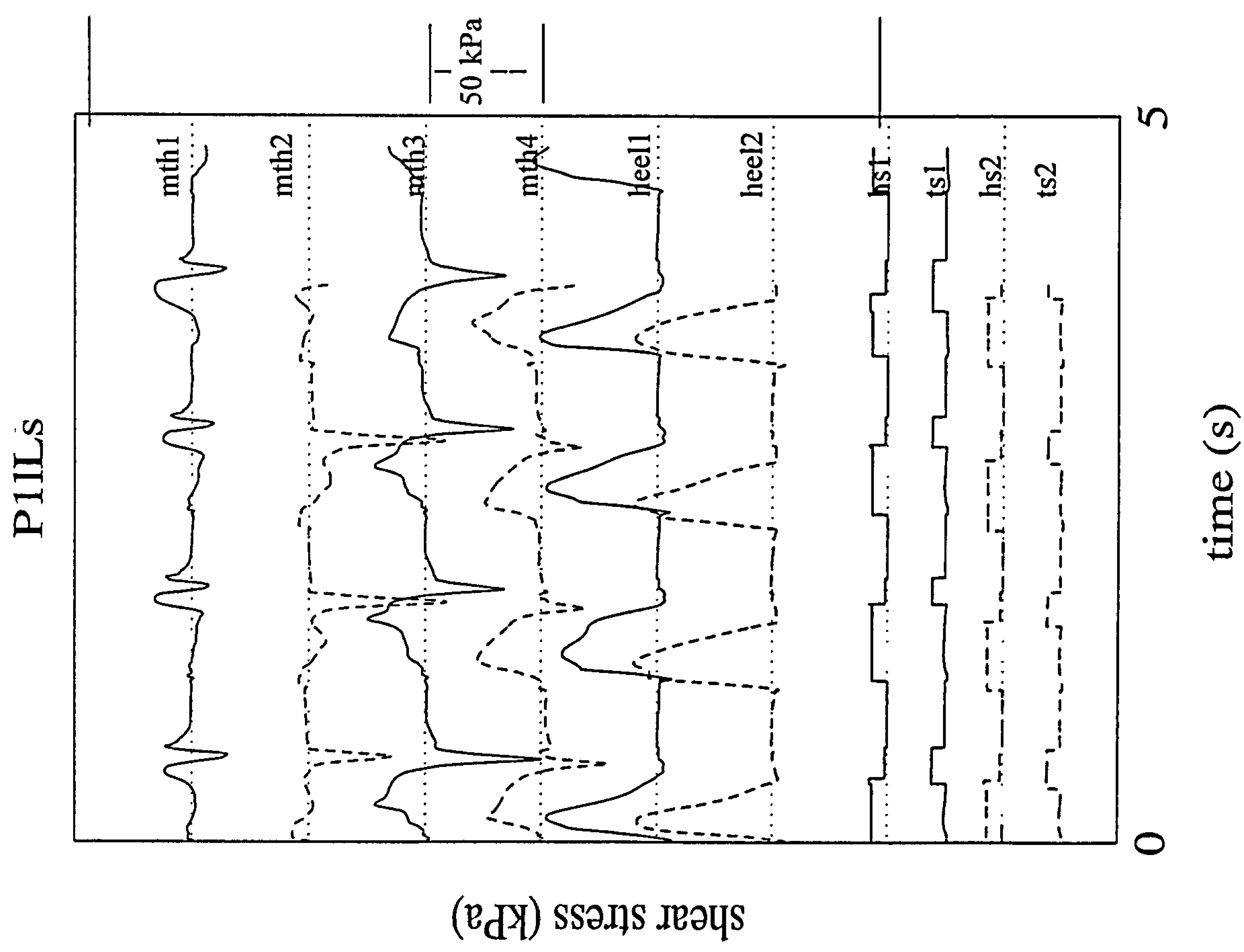

P71Rb-R
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

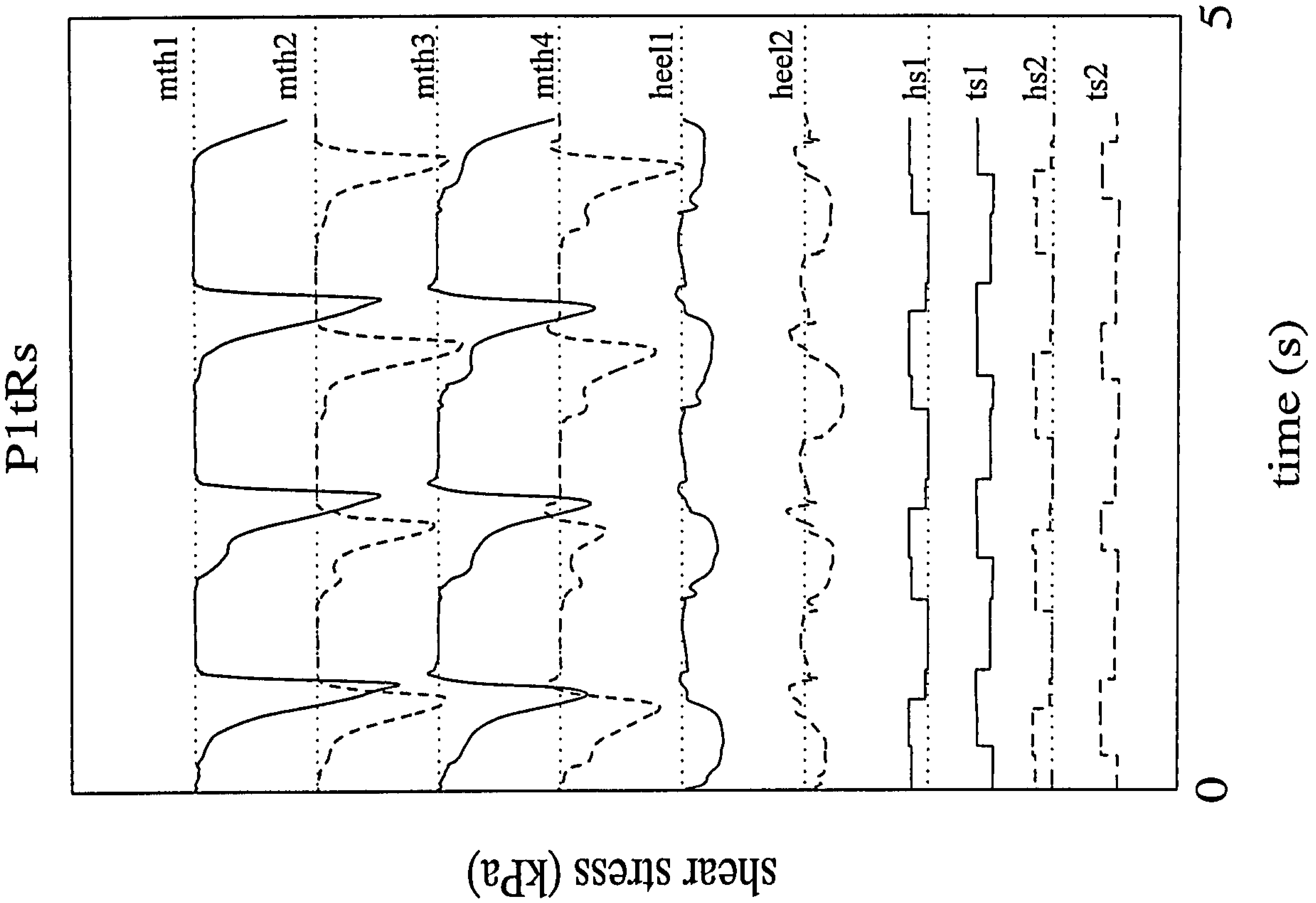

P71Lb-R
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

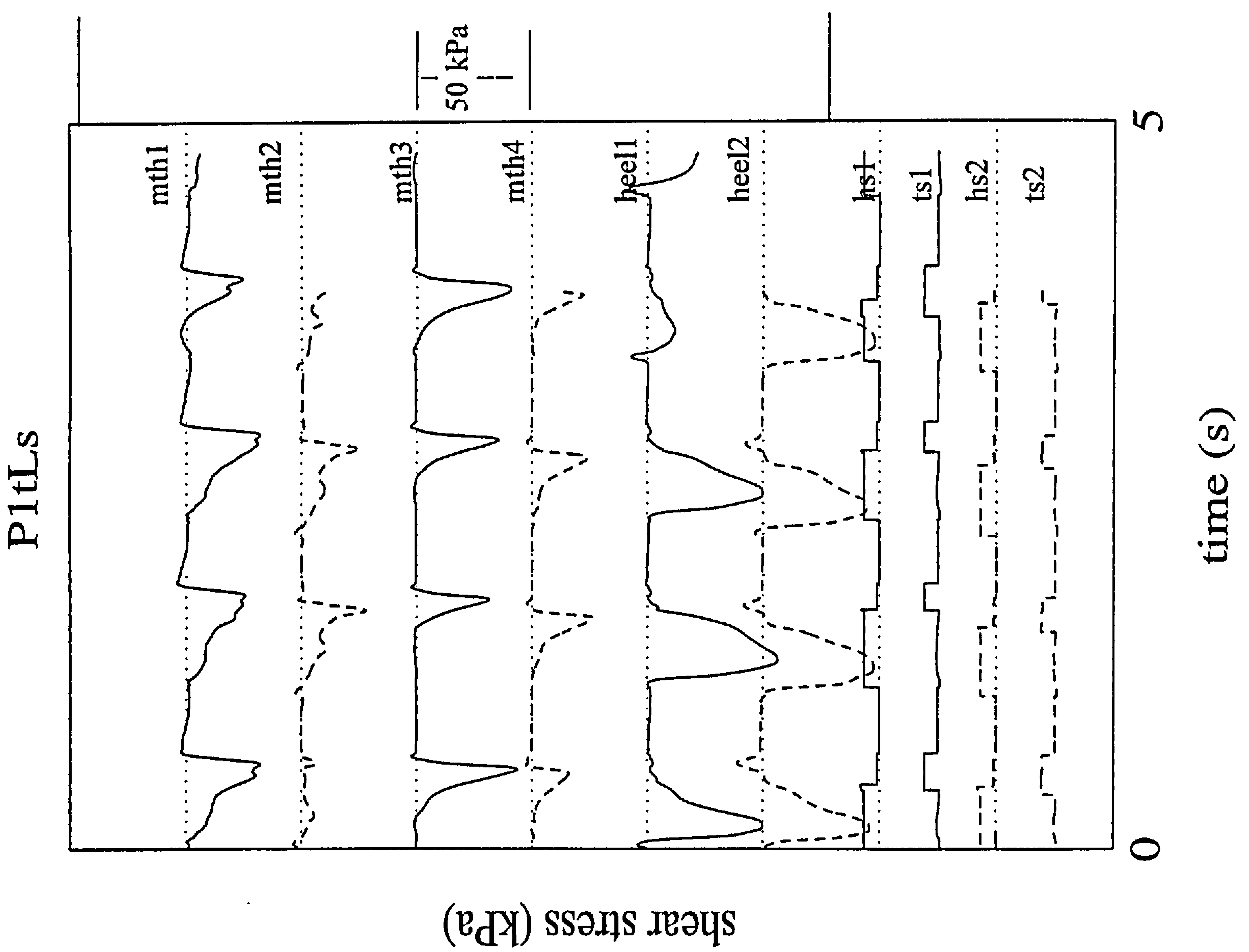

P7tRb-R
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

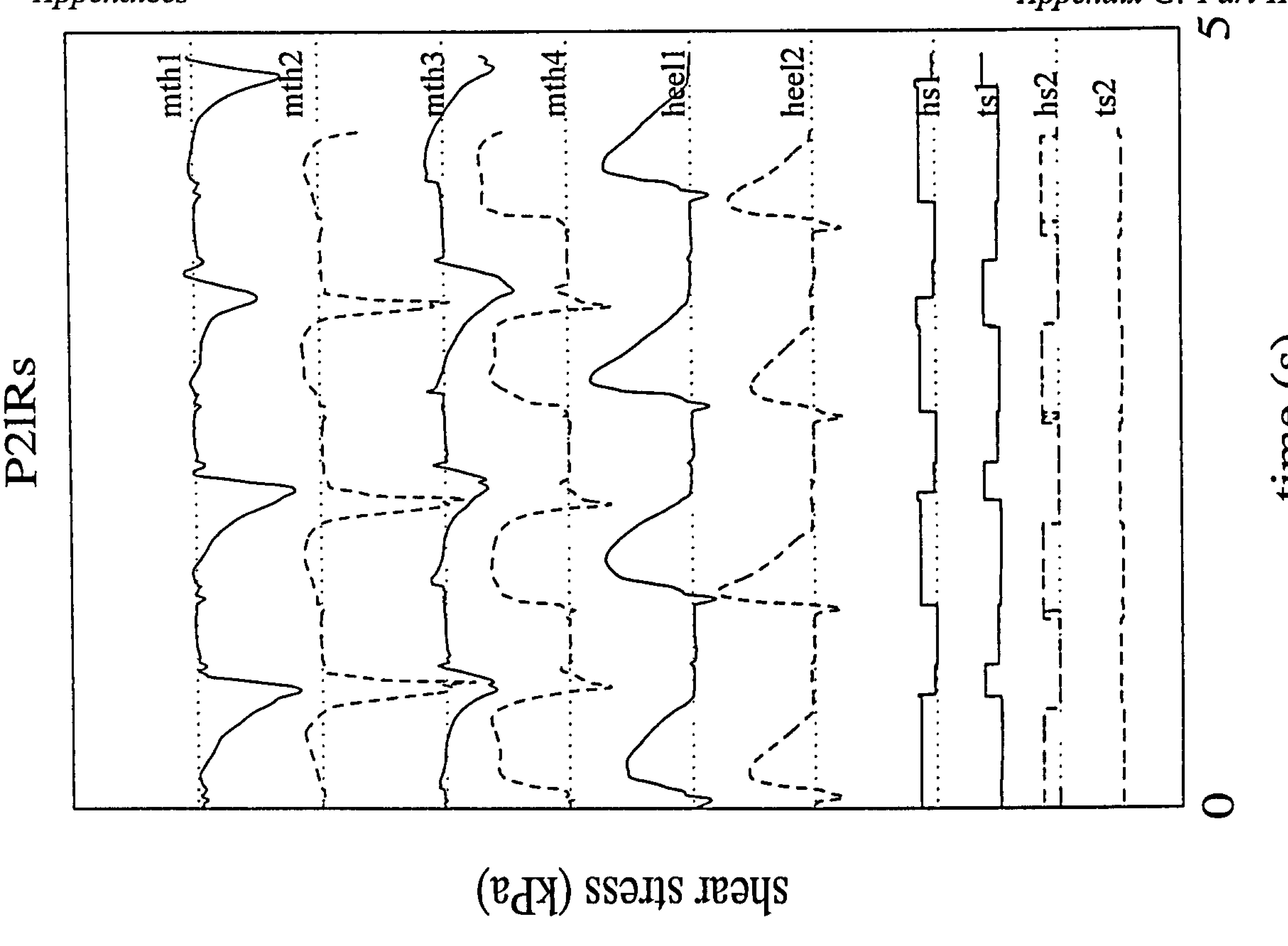

P7tLb-R
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

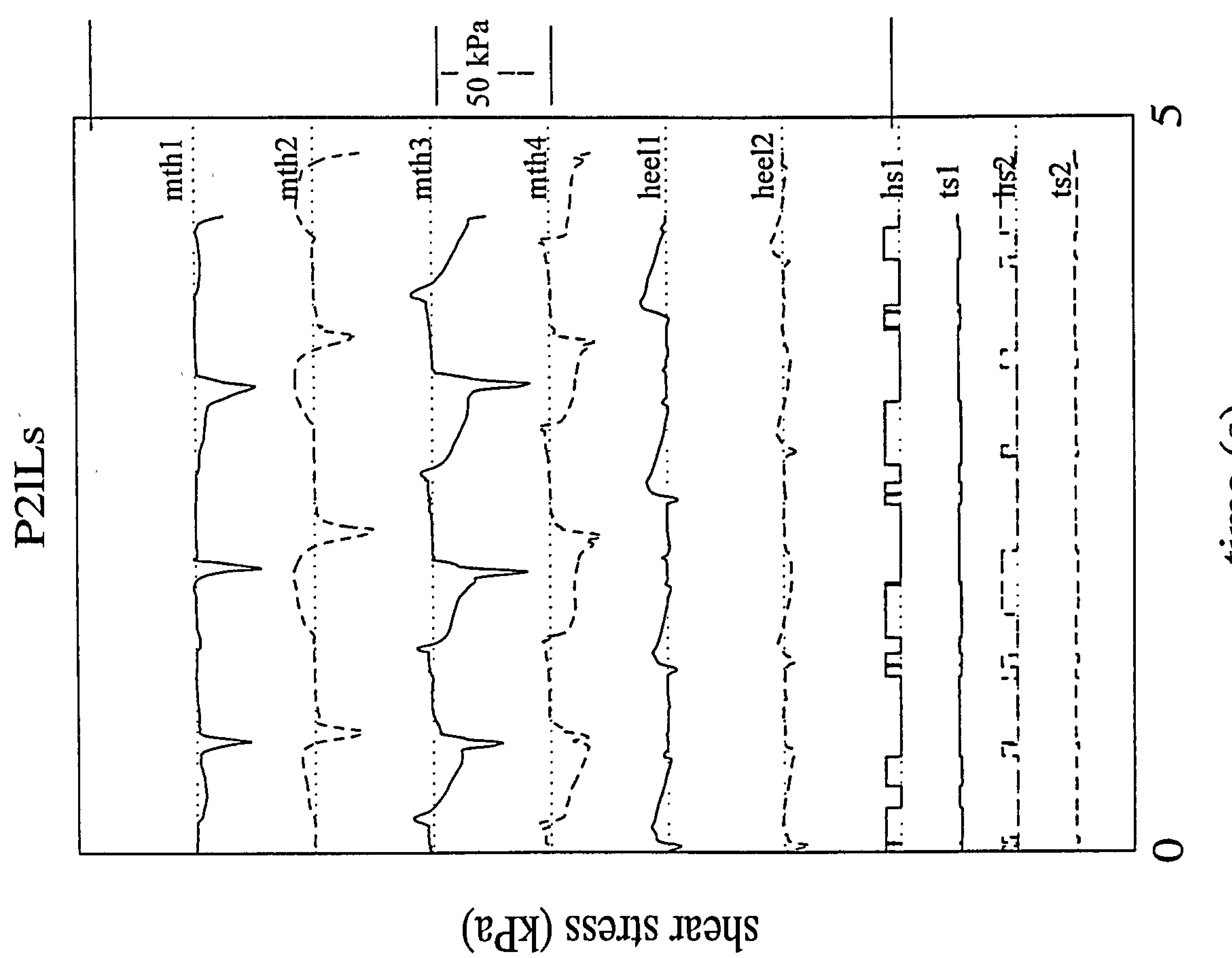

P11Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

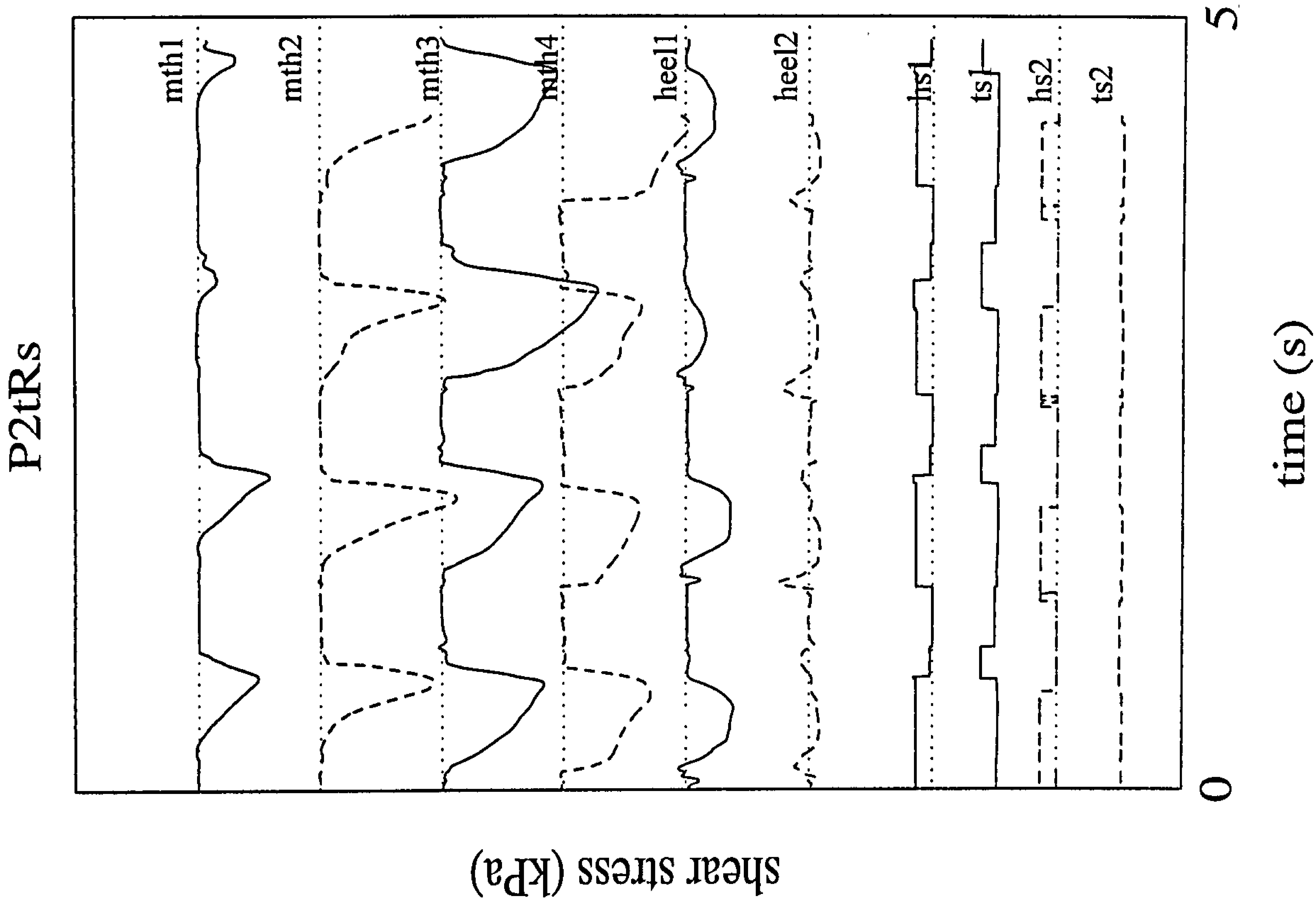

P11Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

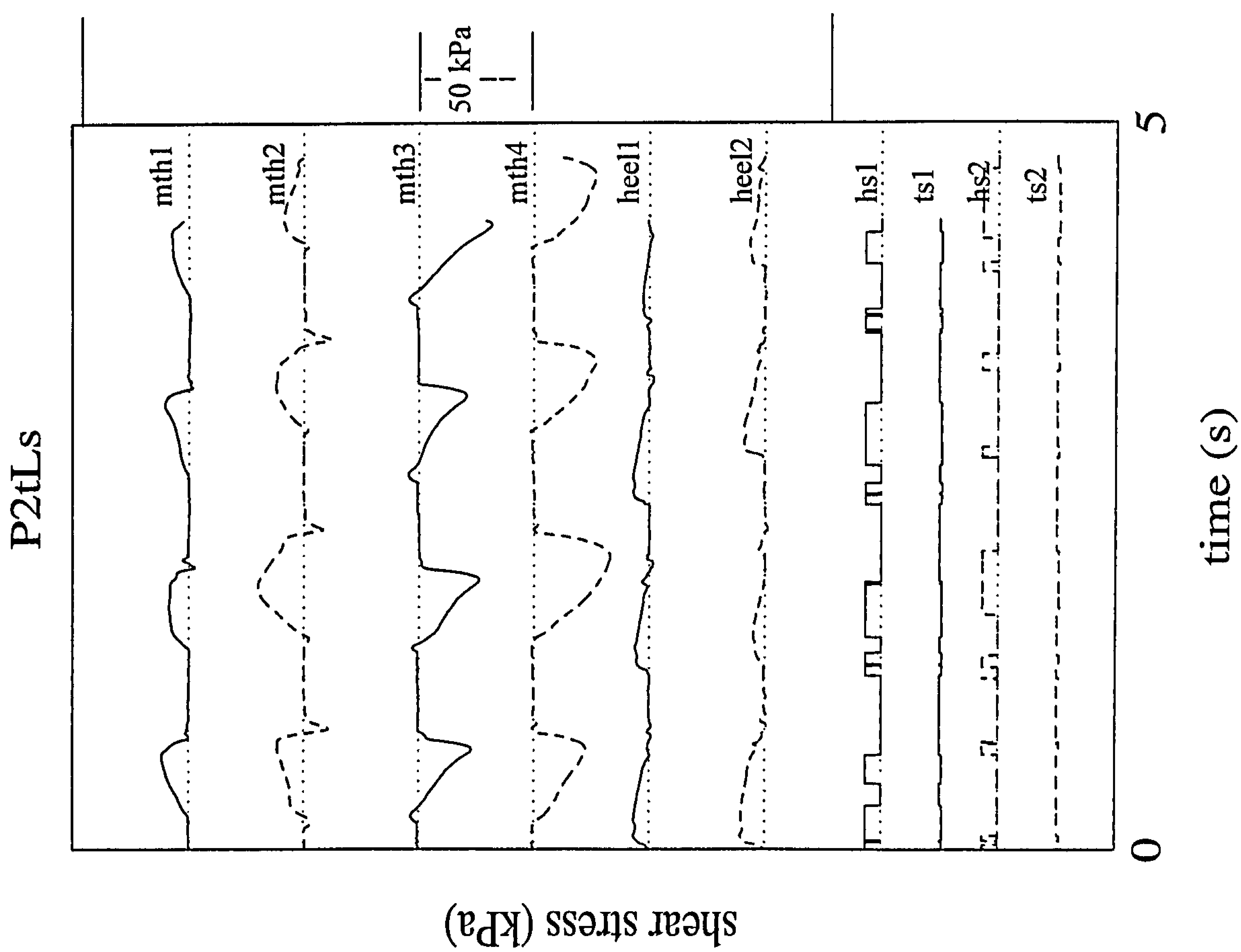

P1tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

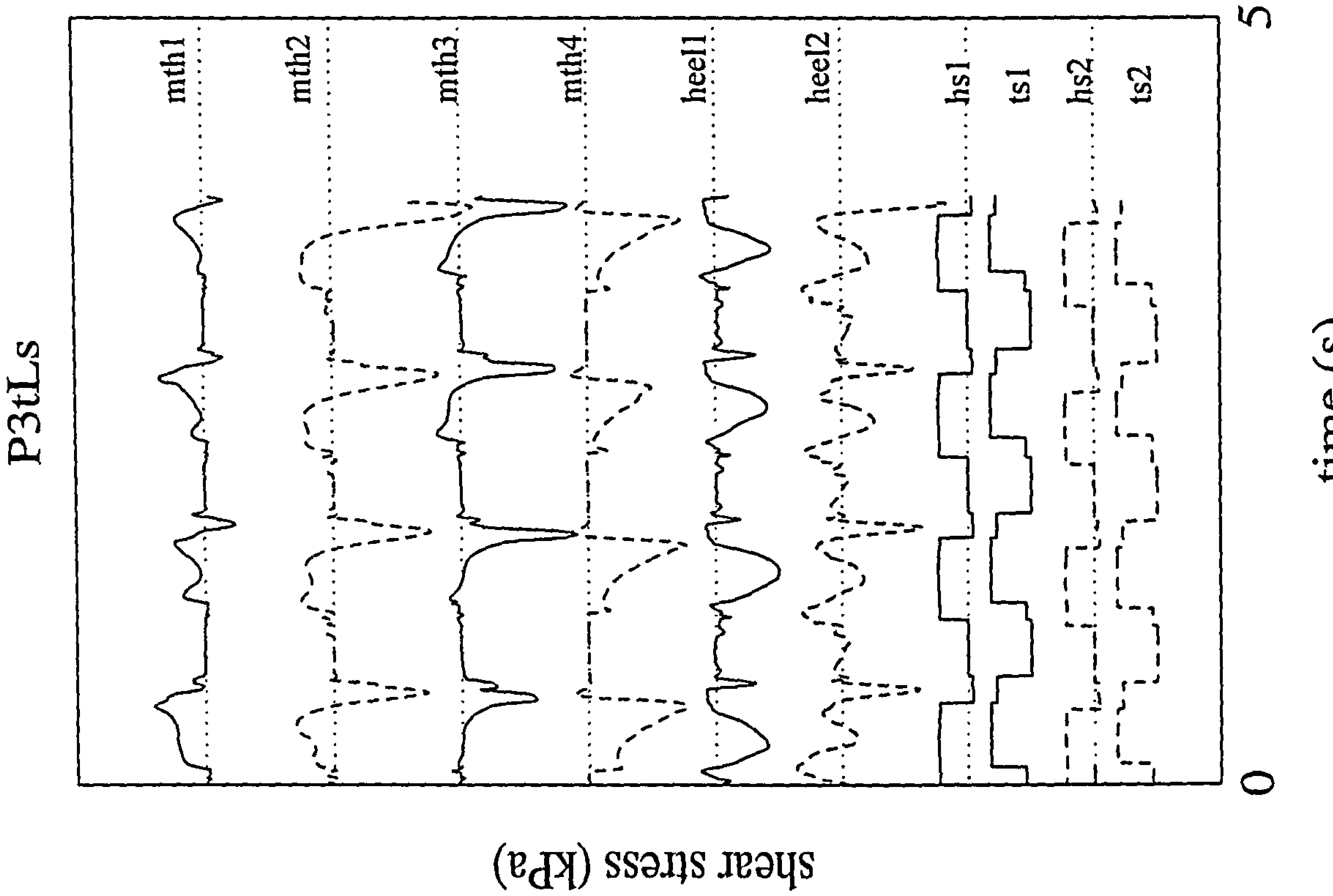

P1tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

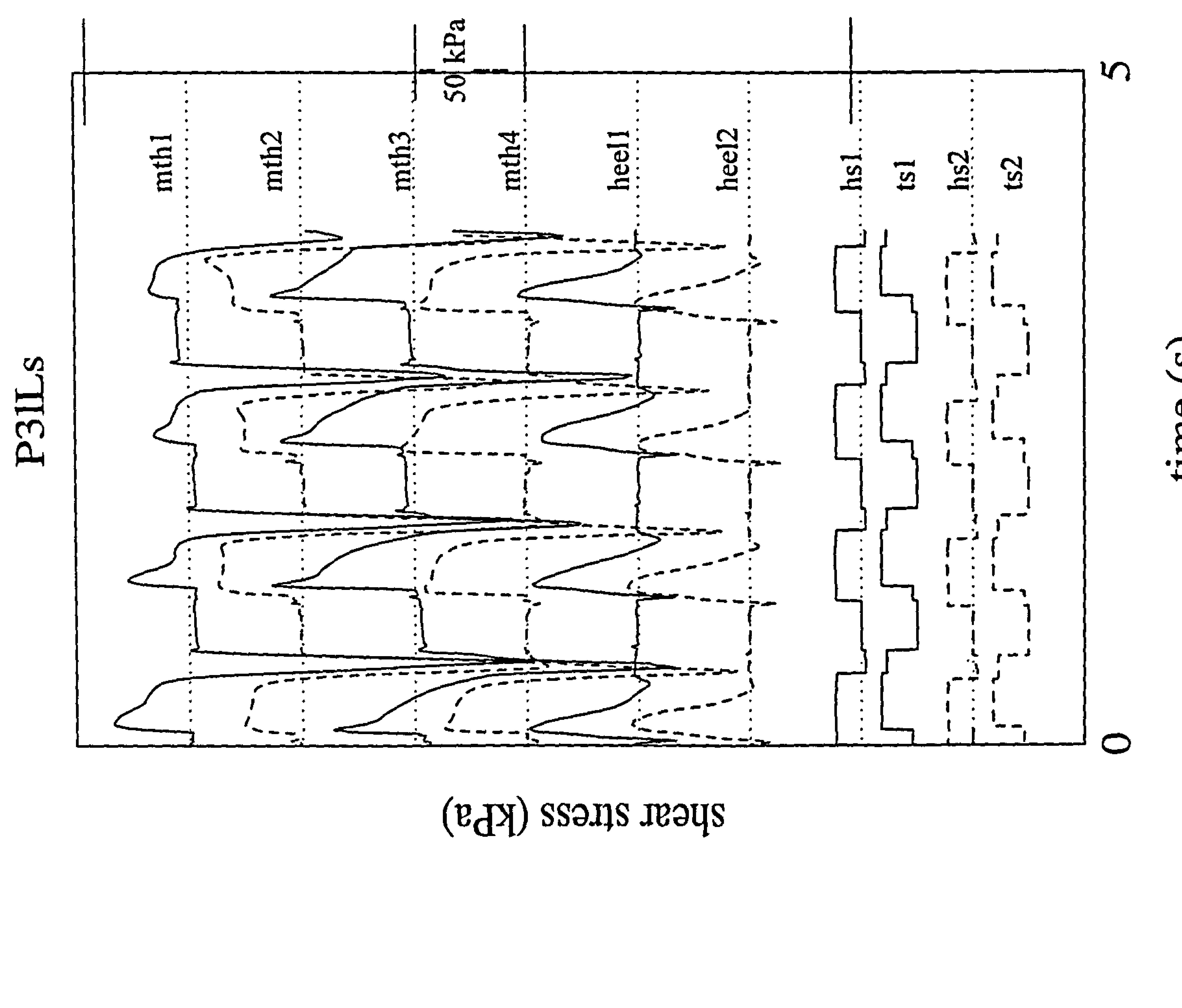

P21Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

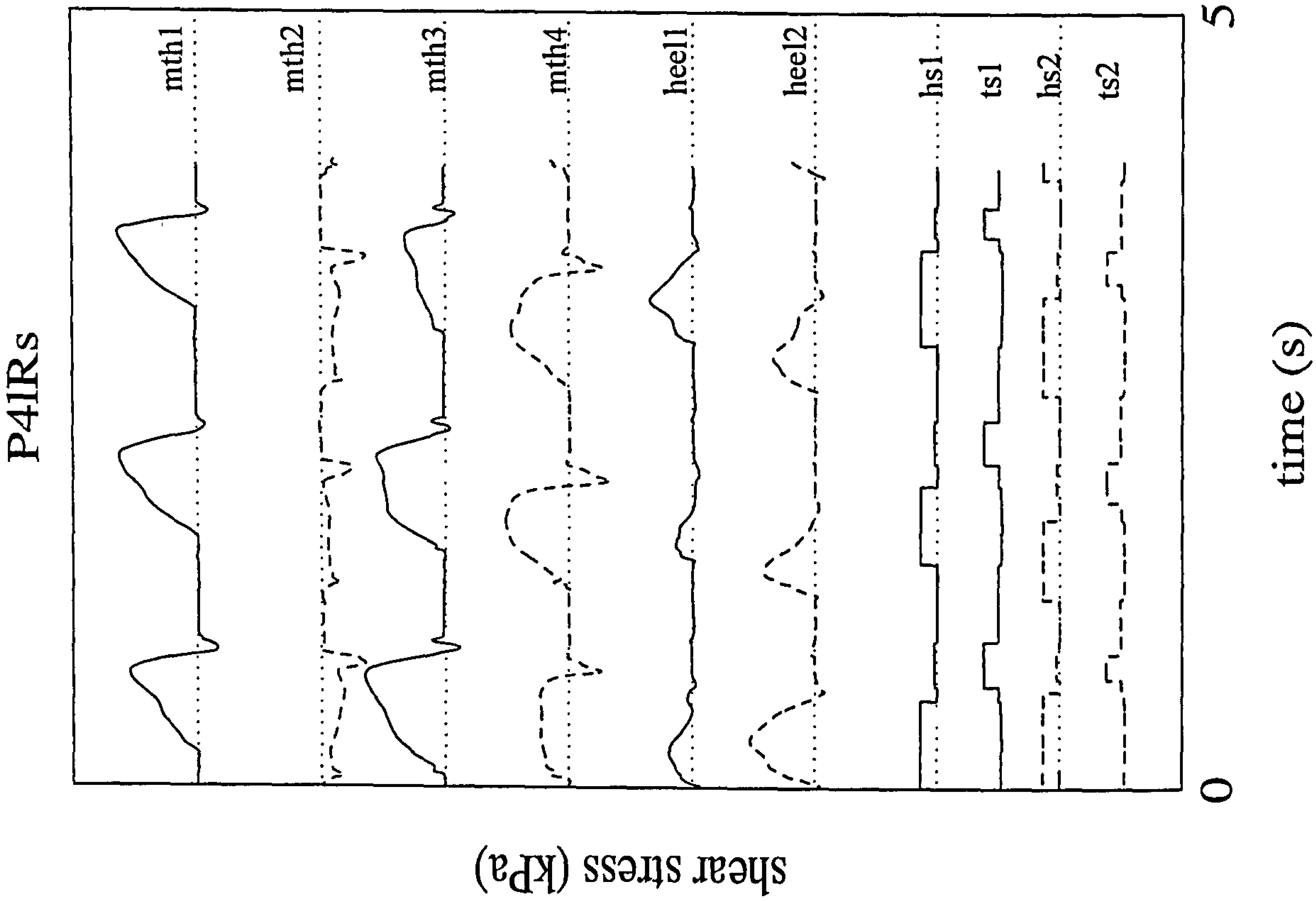

P21Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

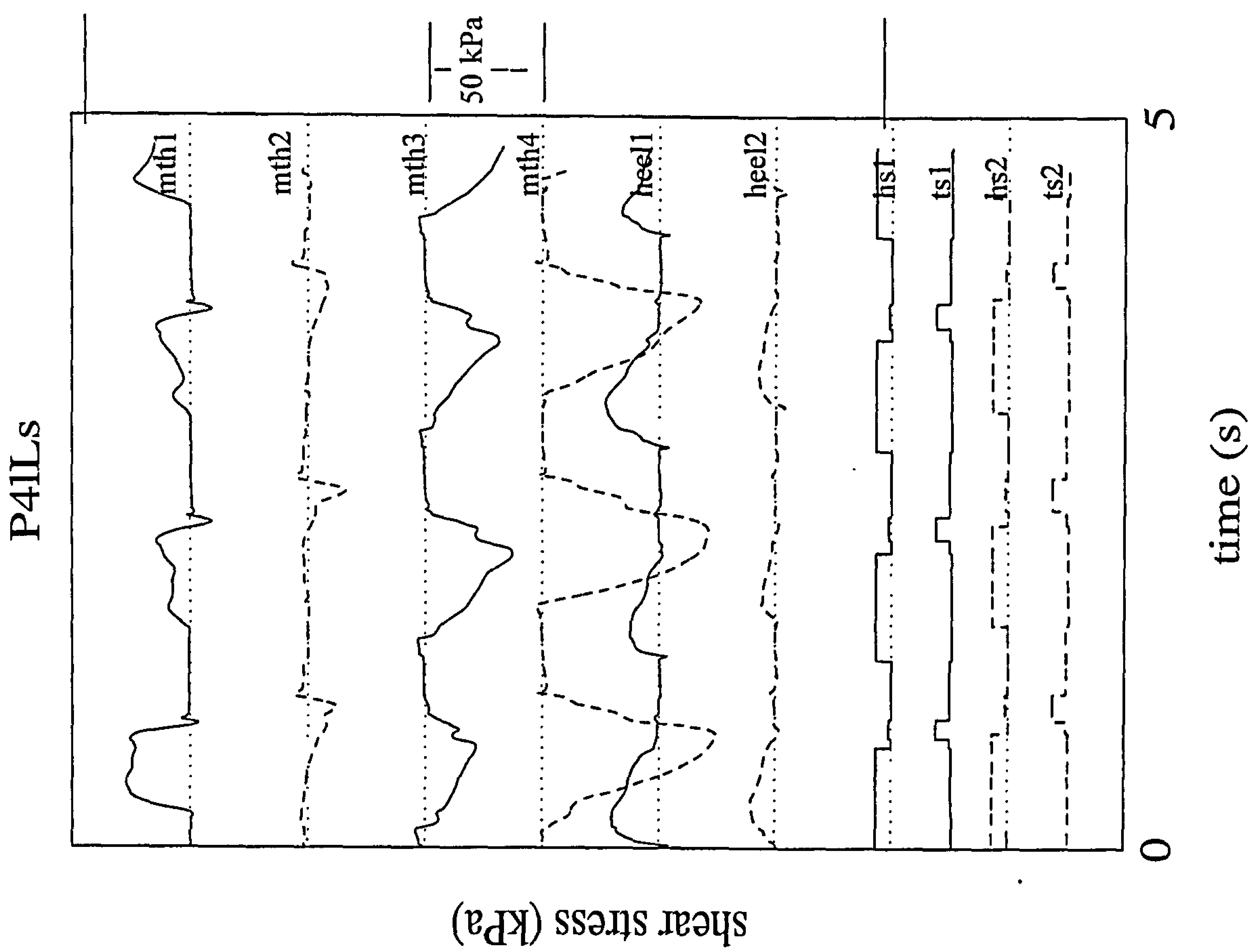

P2tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

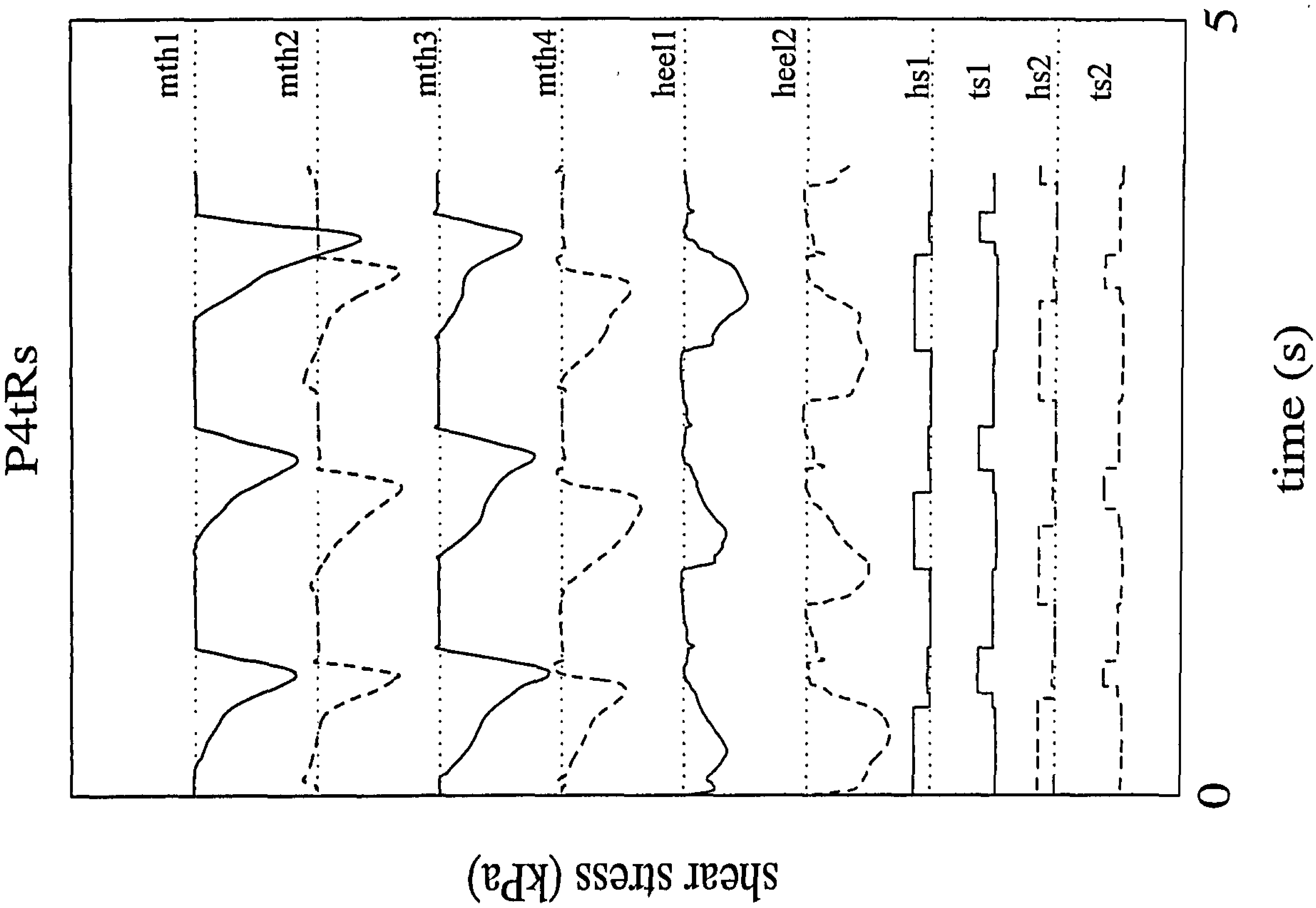

P2tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

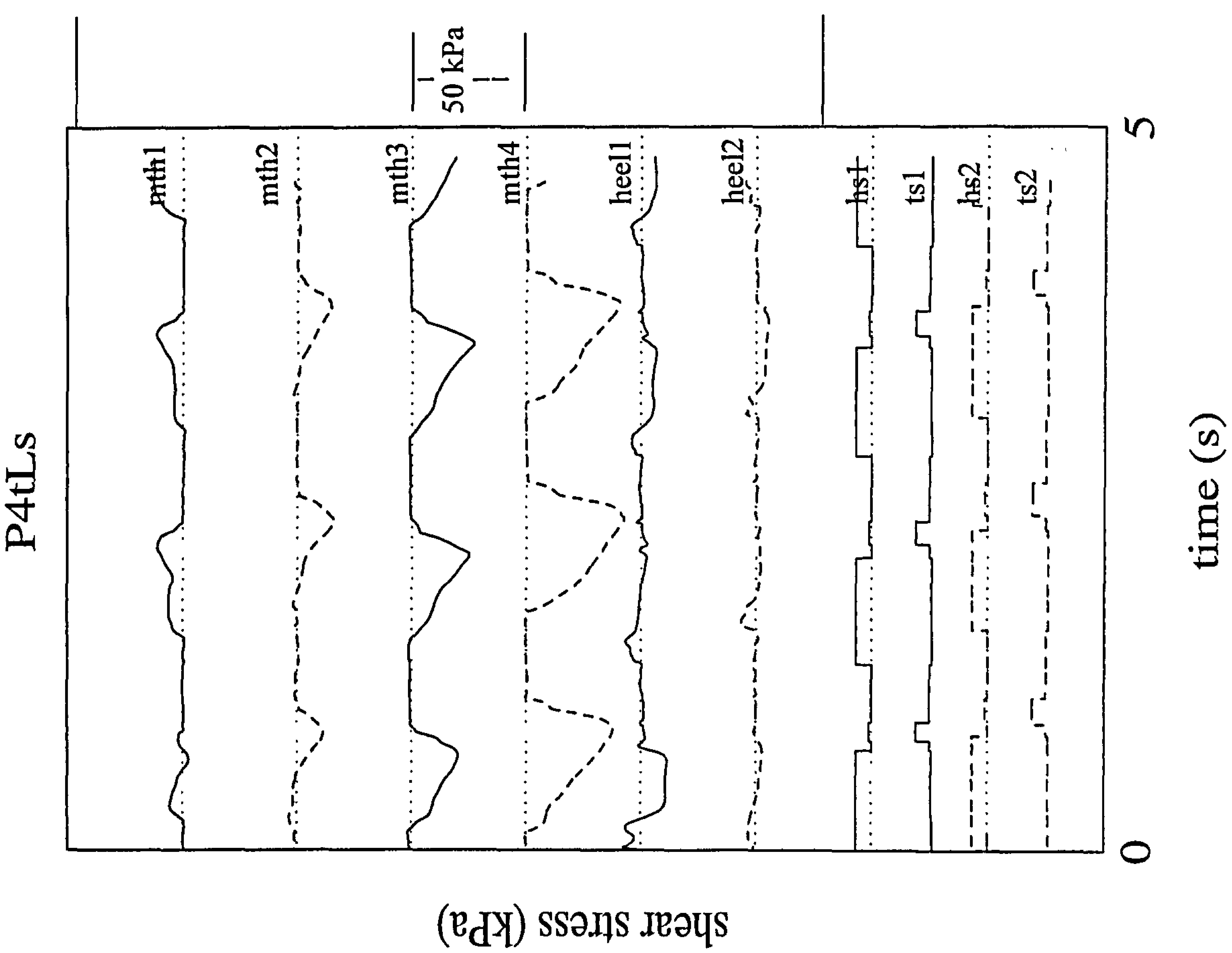

P3tLs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

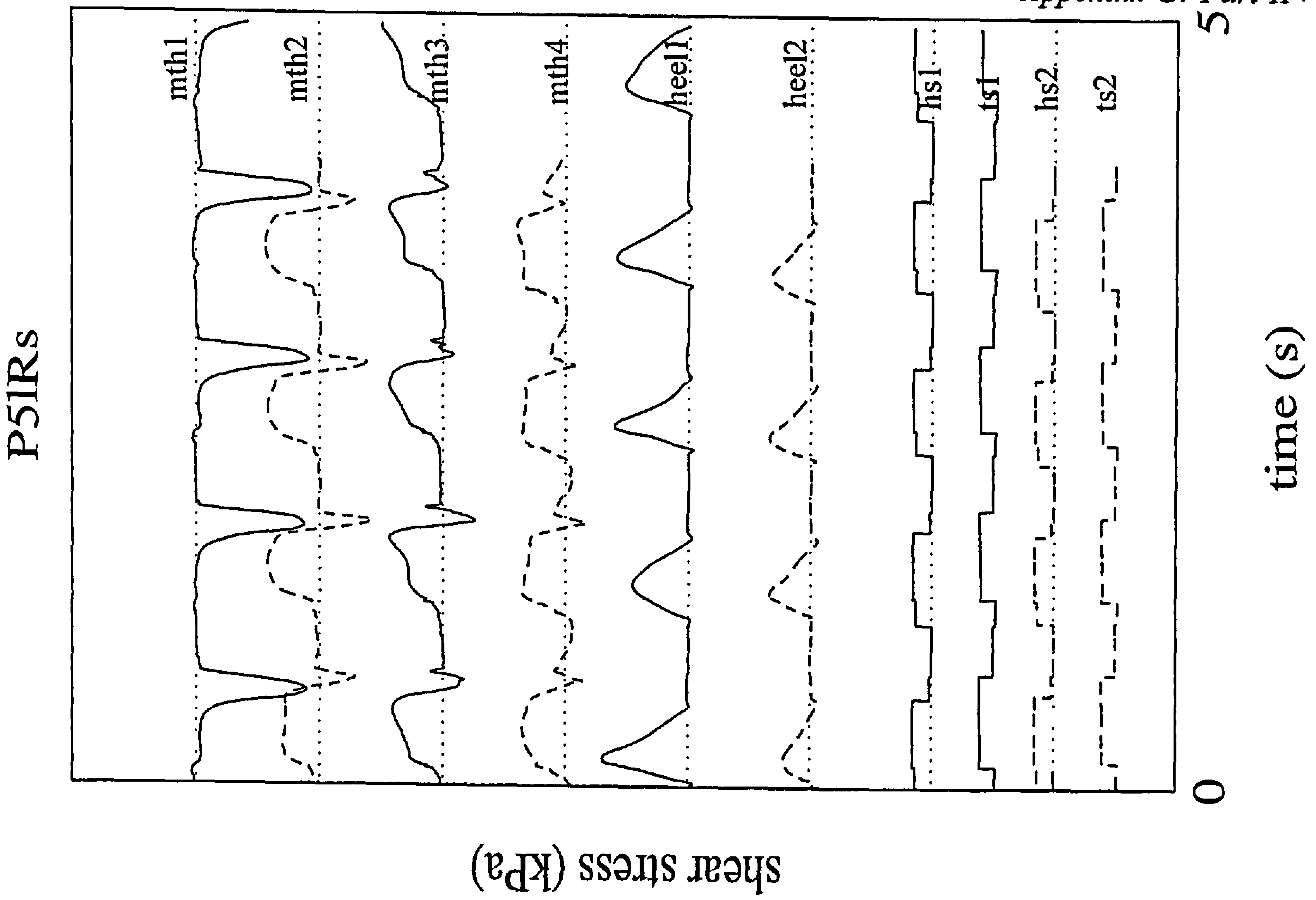

P3lLs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
50 kPa
shear stress (kPa)
time (s)
0
5

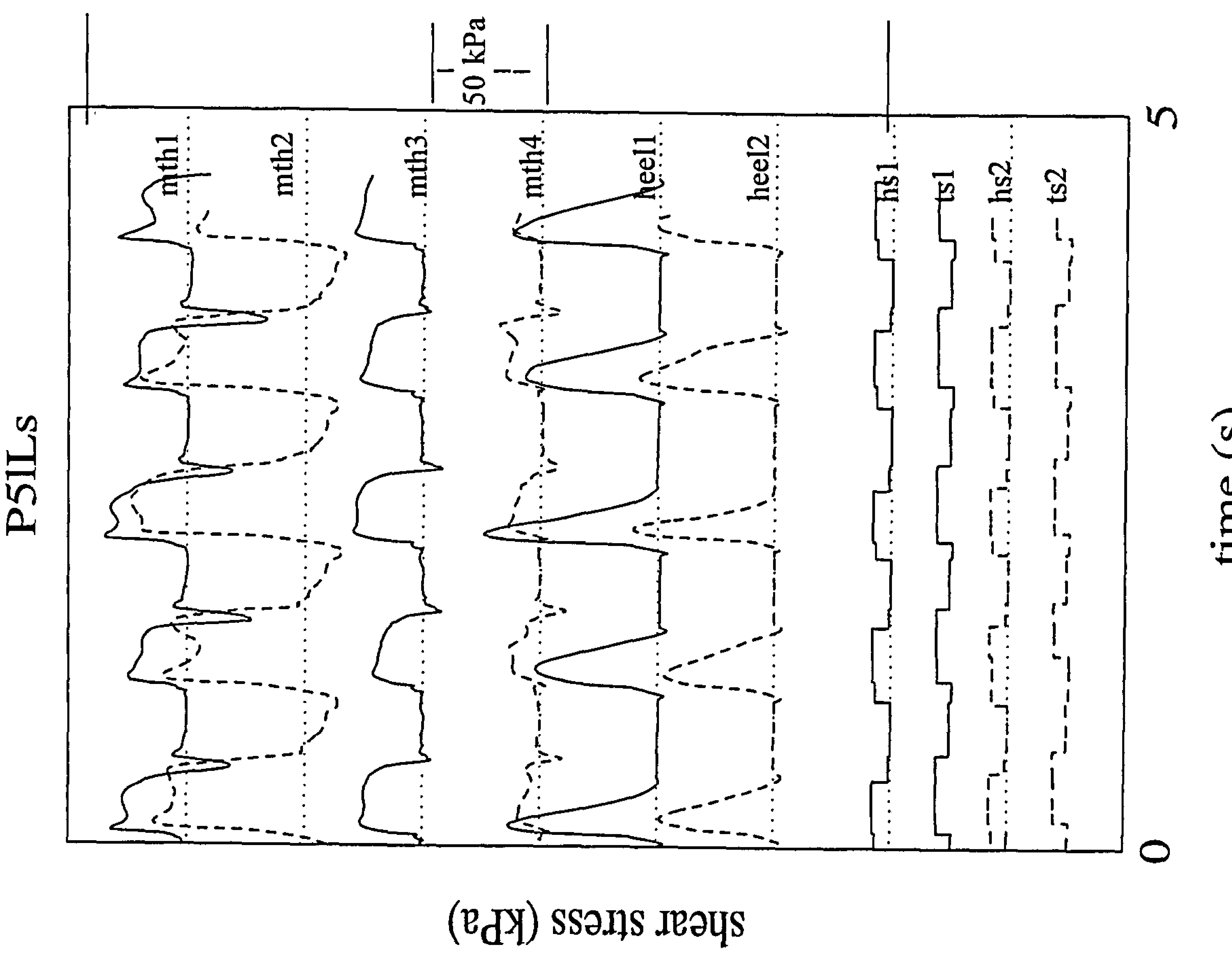

P41Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

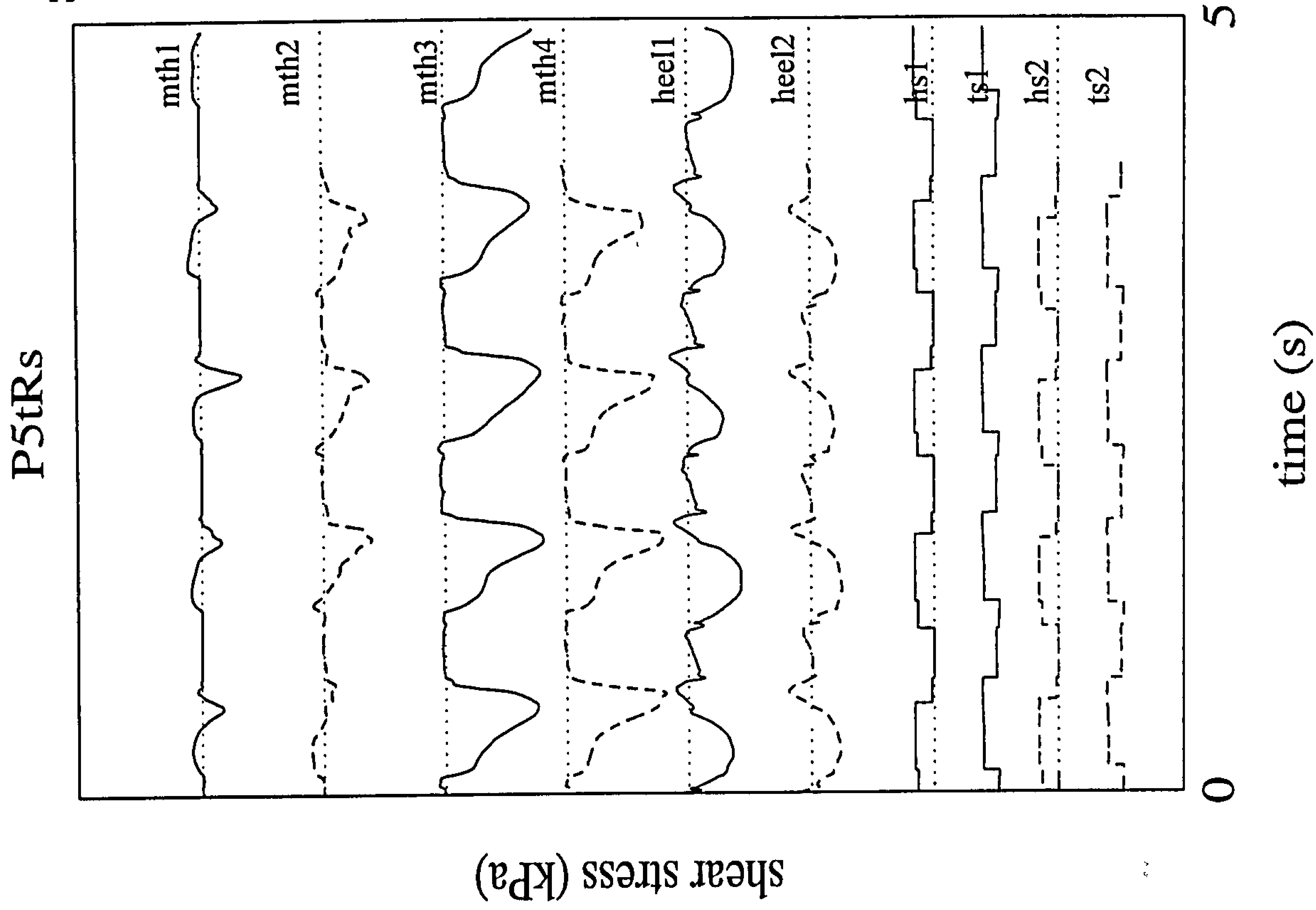

P41Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

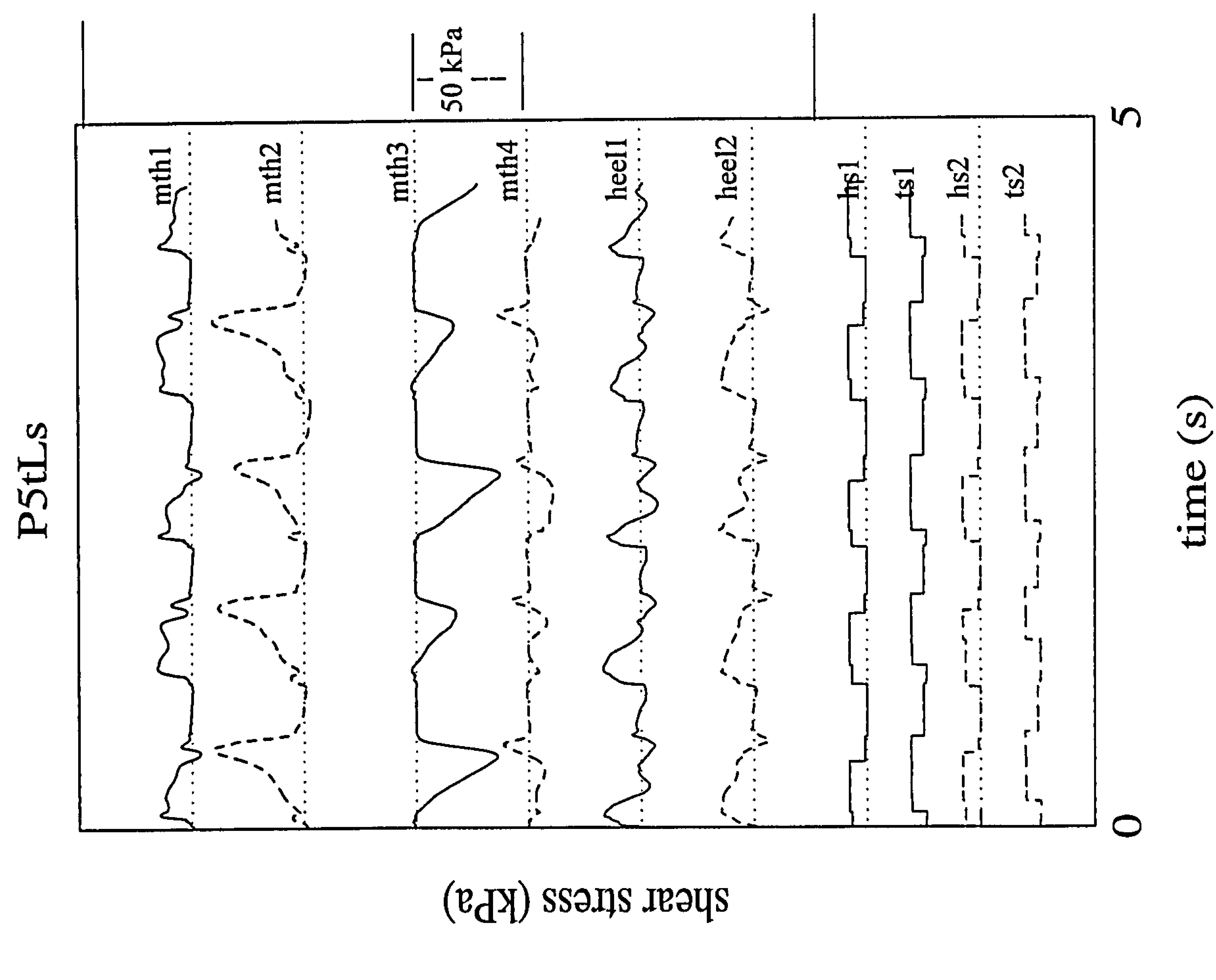

P4tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

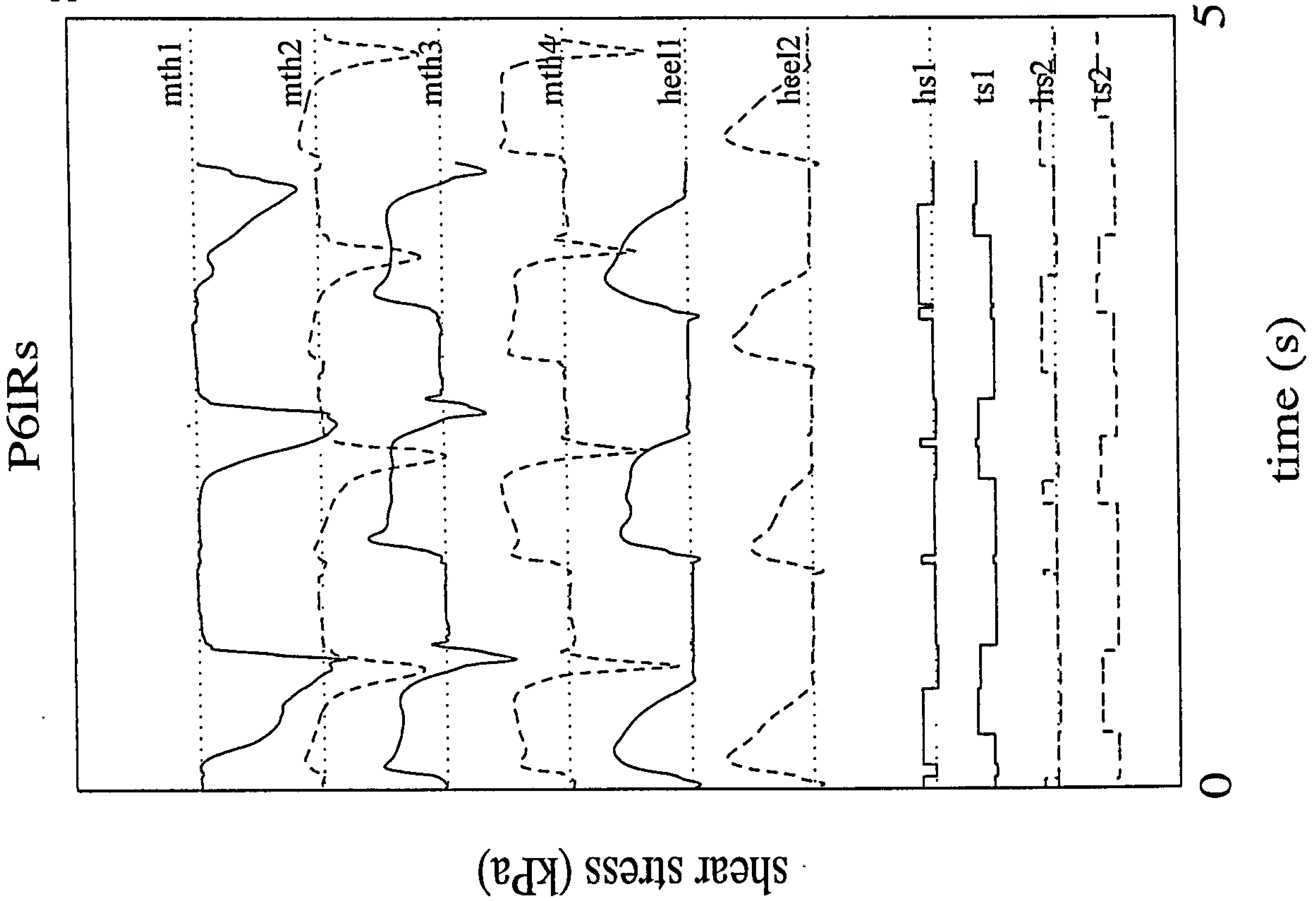

P4tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


P51Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)

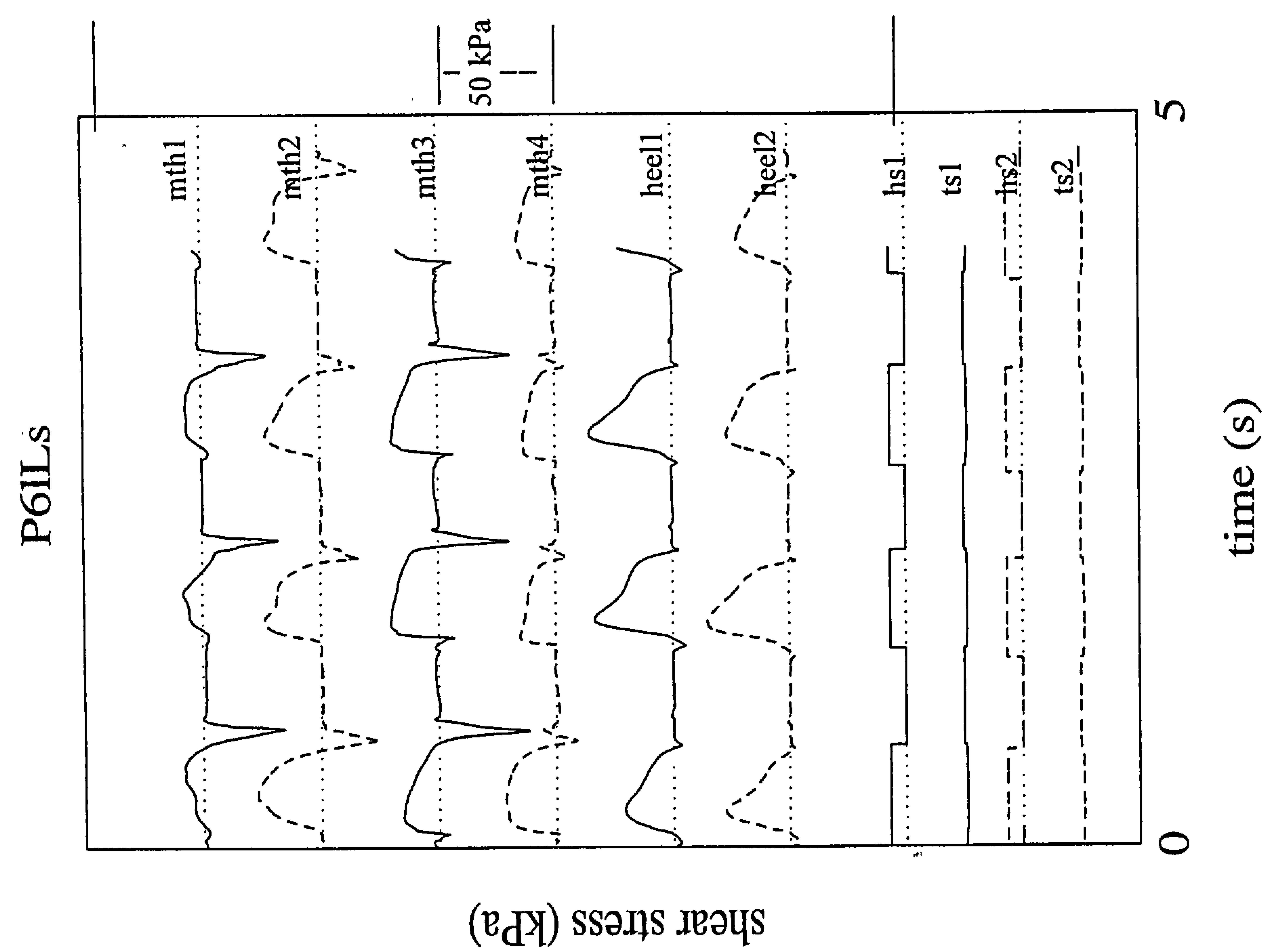

P51Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
0
5
time (s)


P5tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

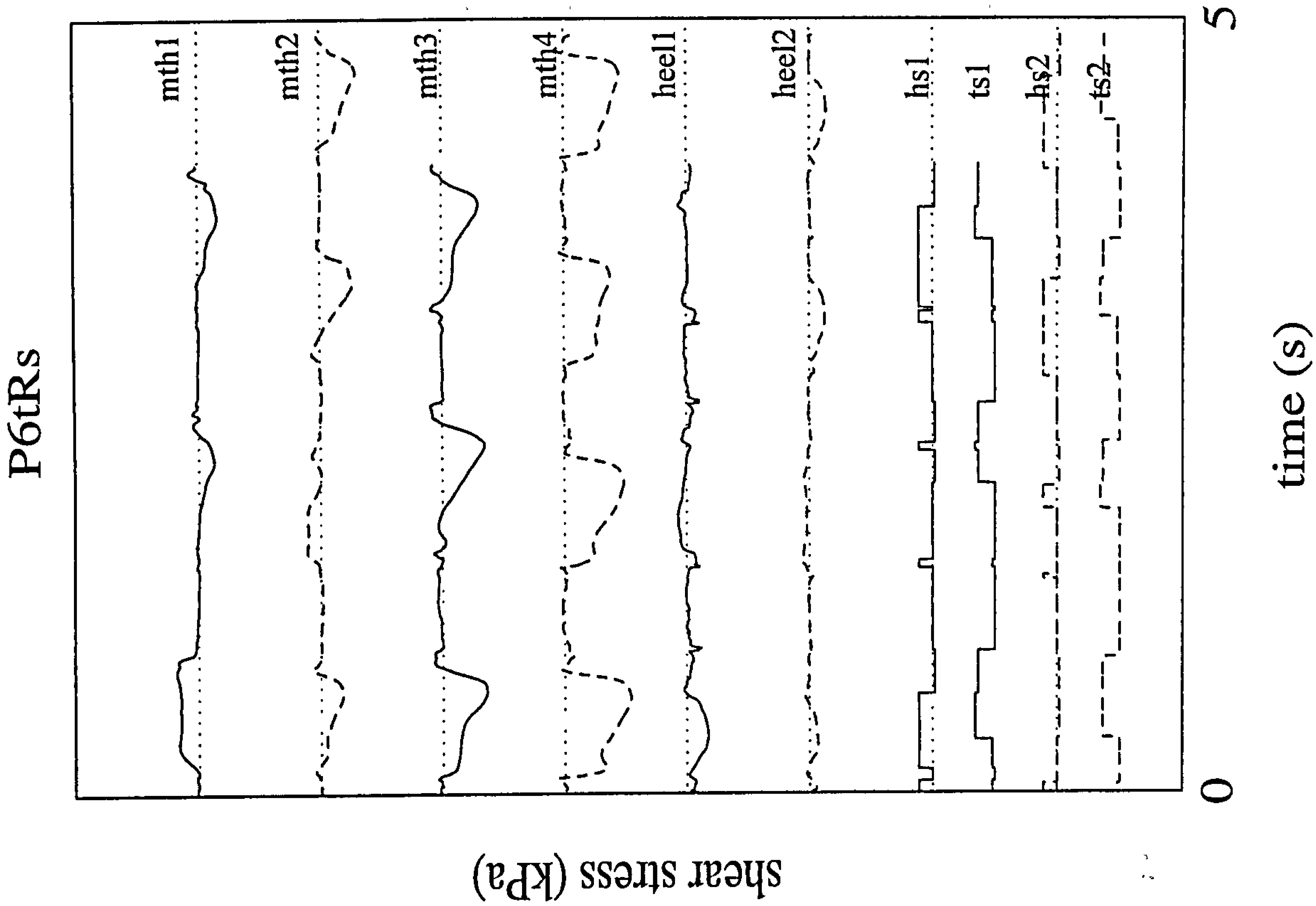

P5tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)



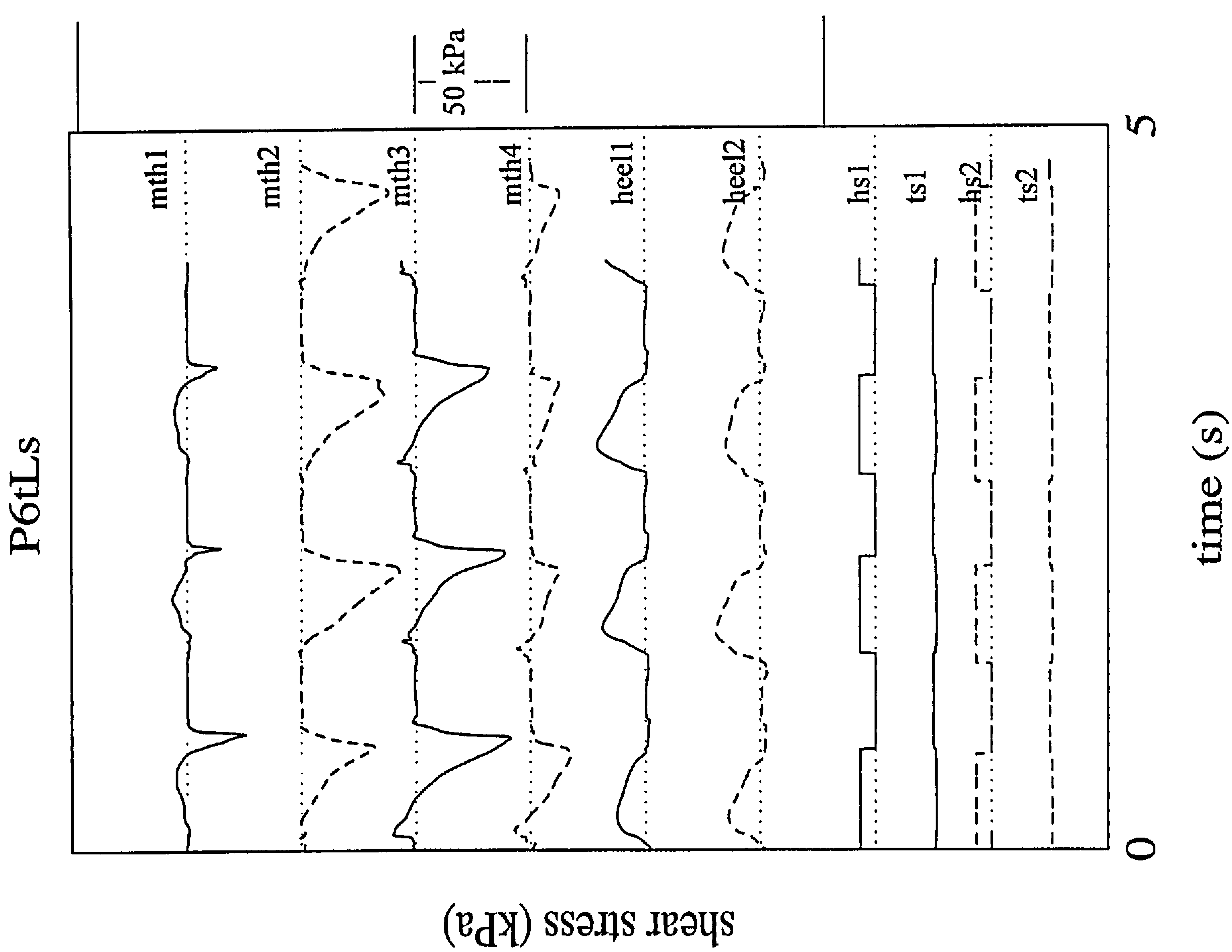
P61Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)



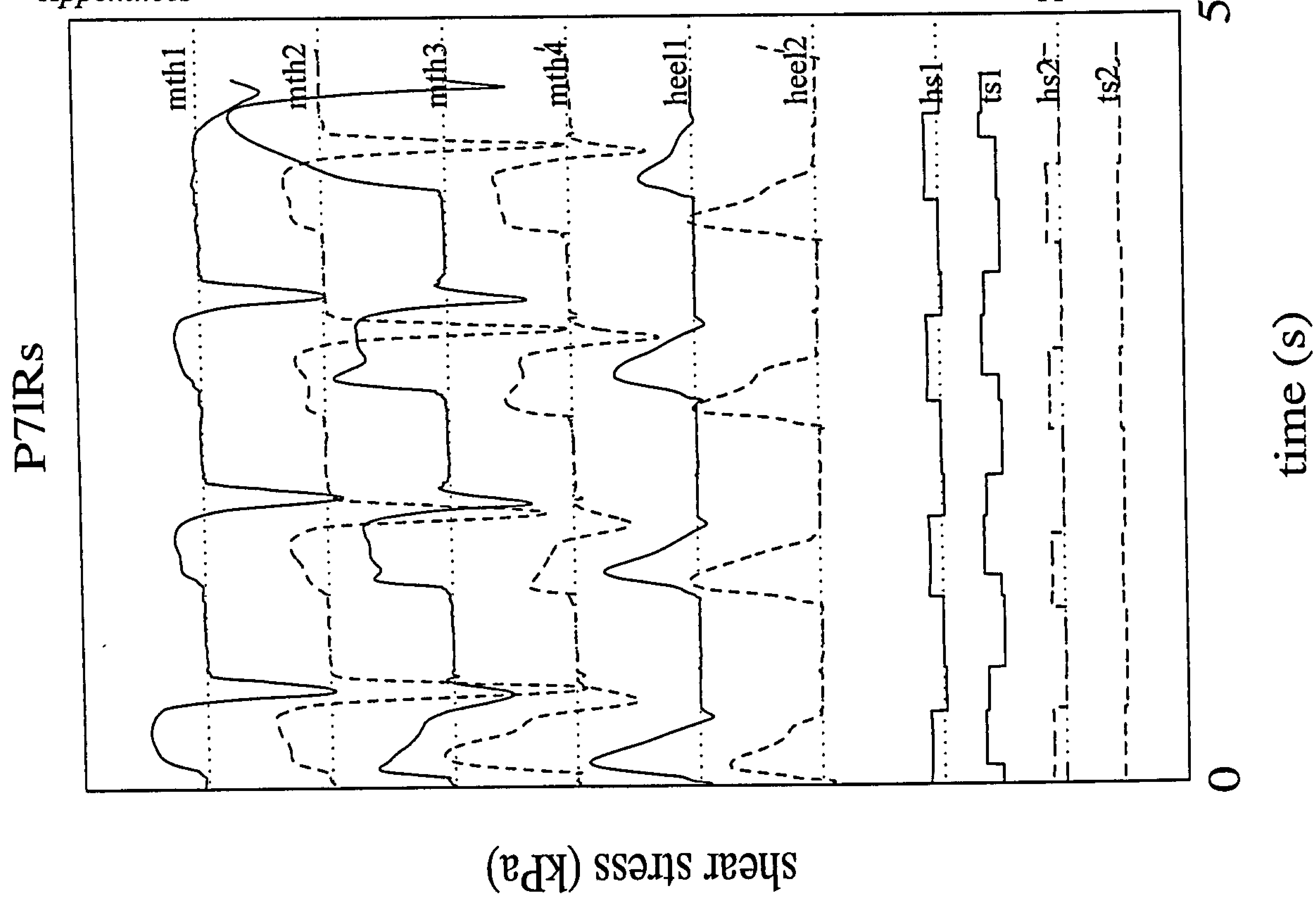
P61Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

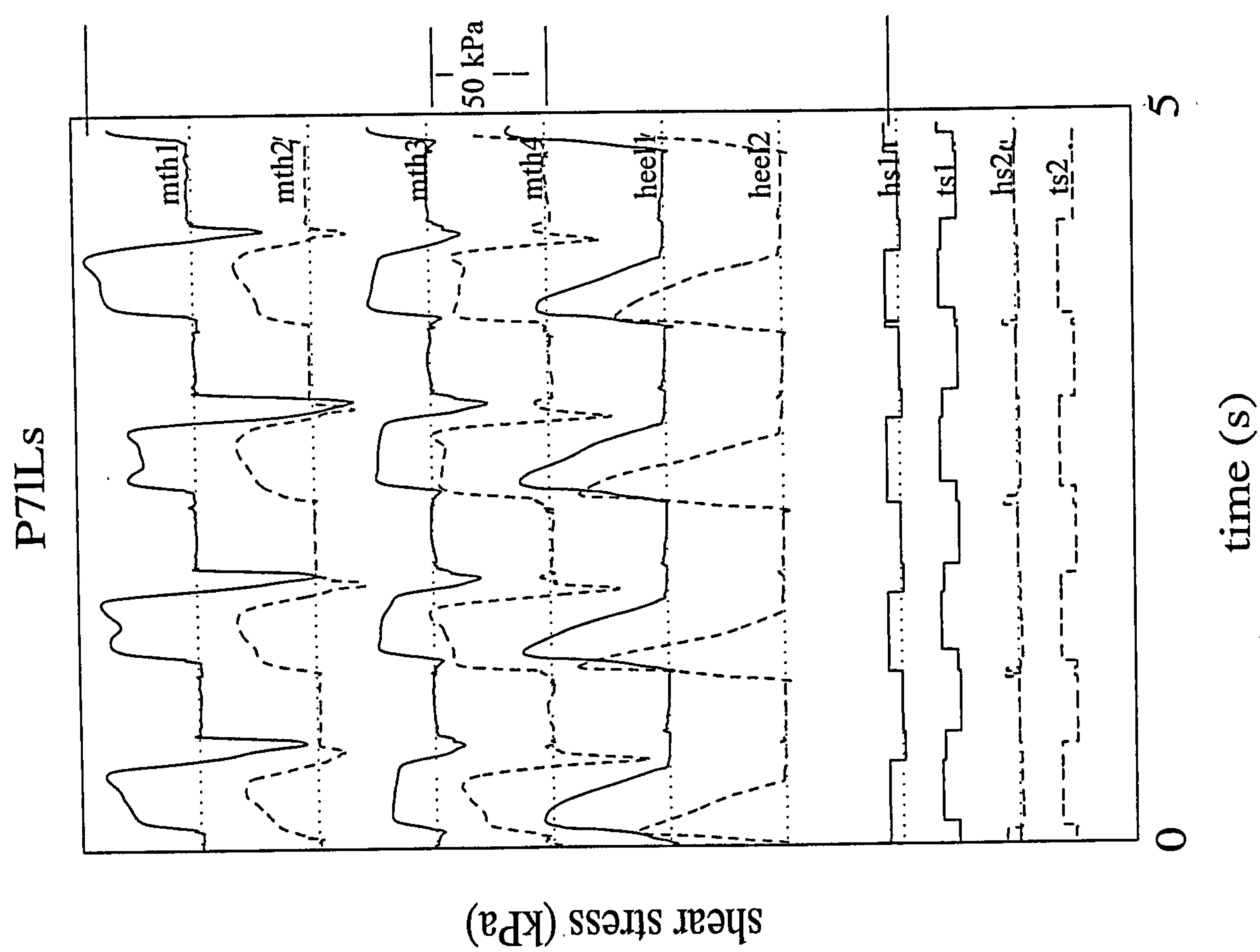


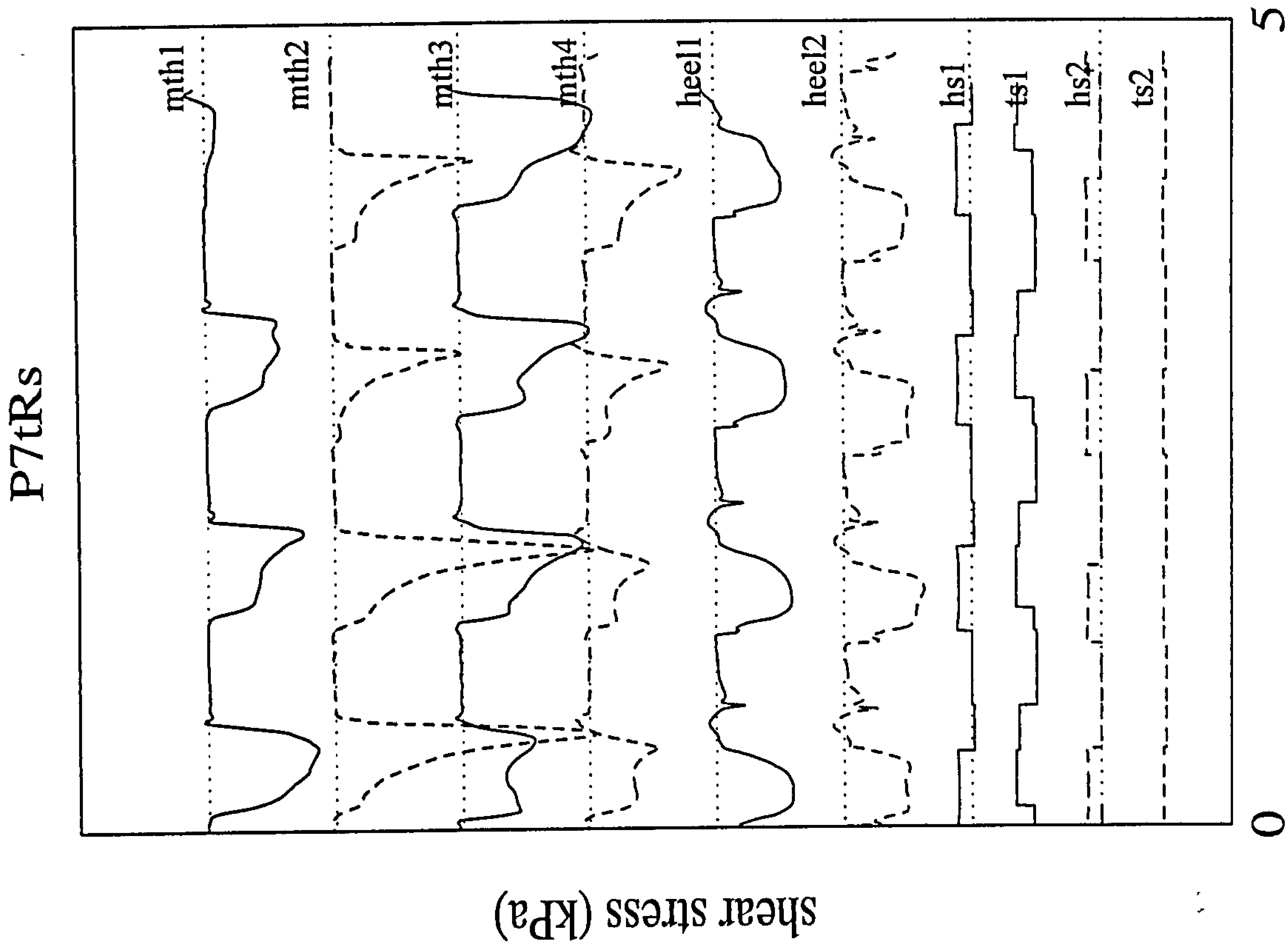
P6tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

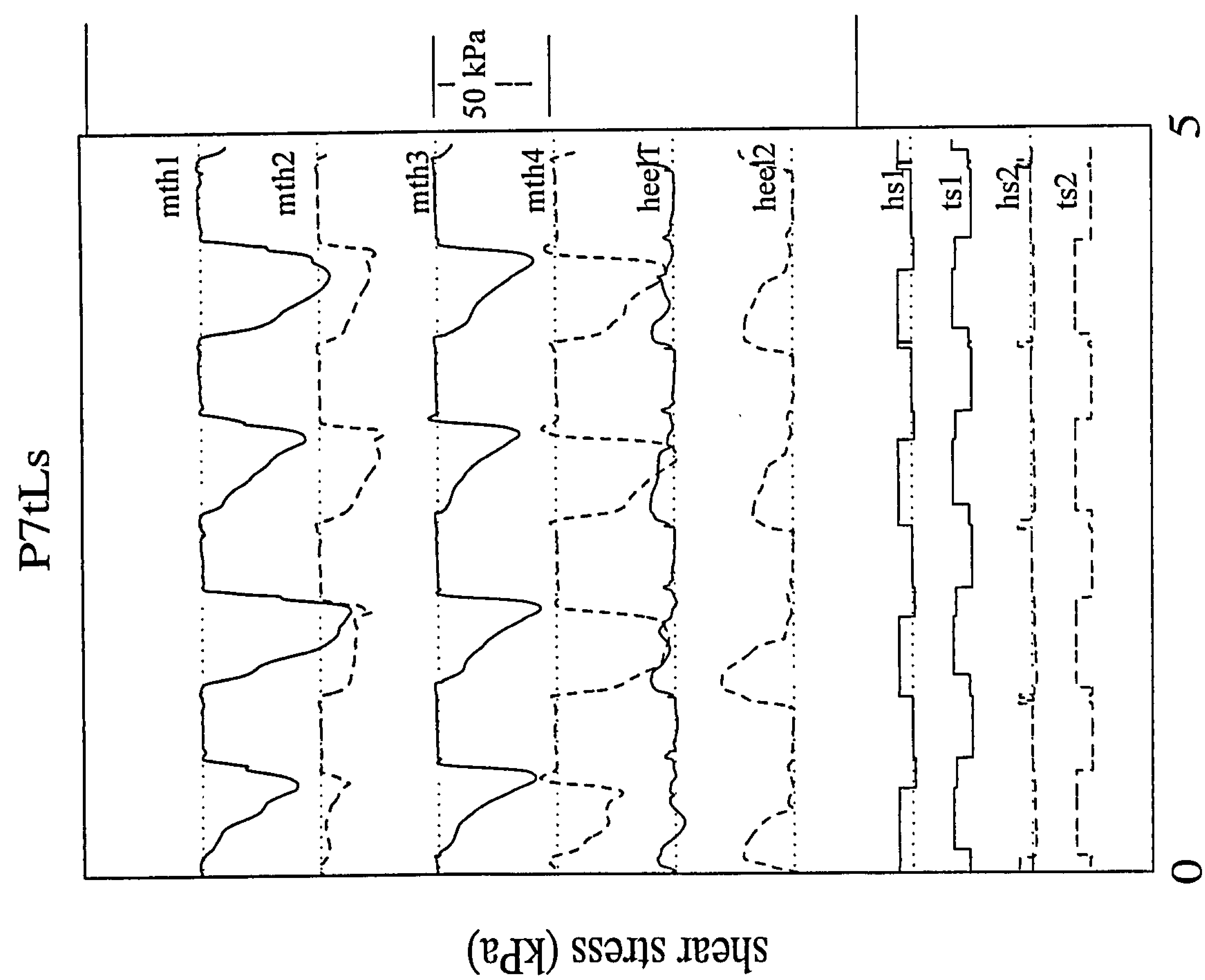


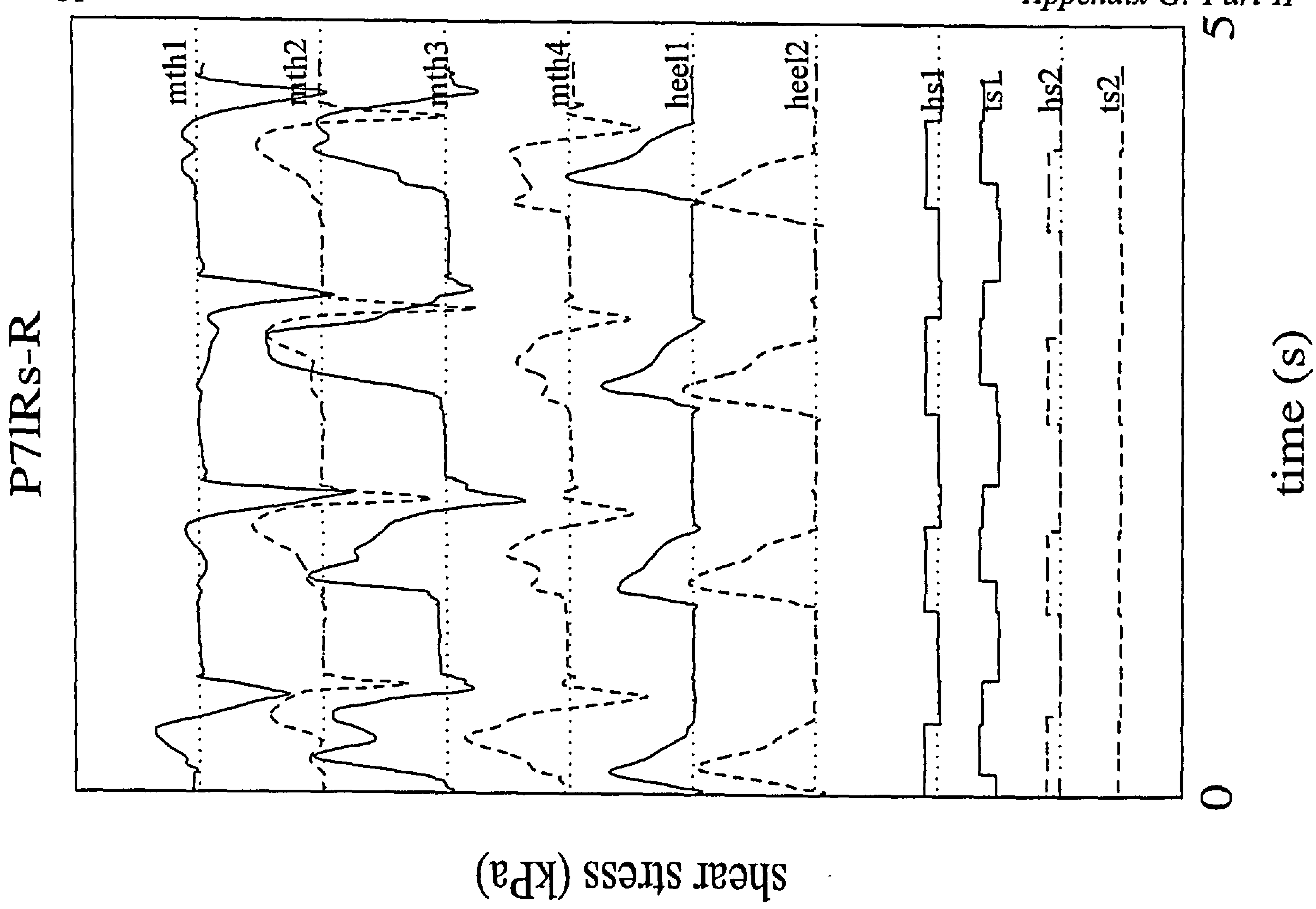
P6tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

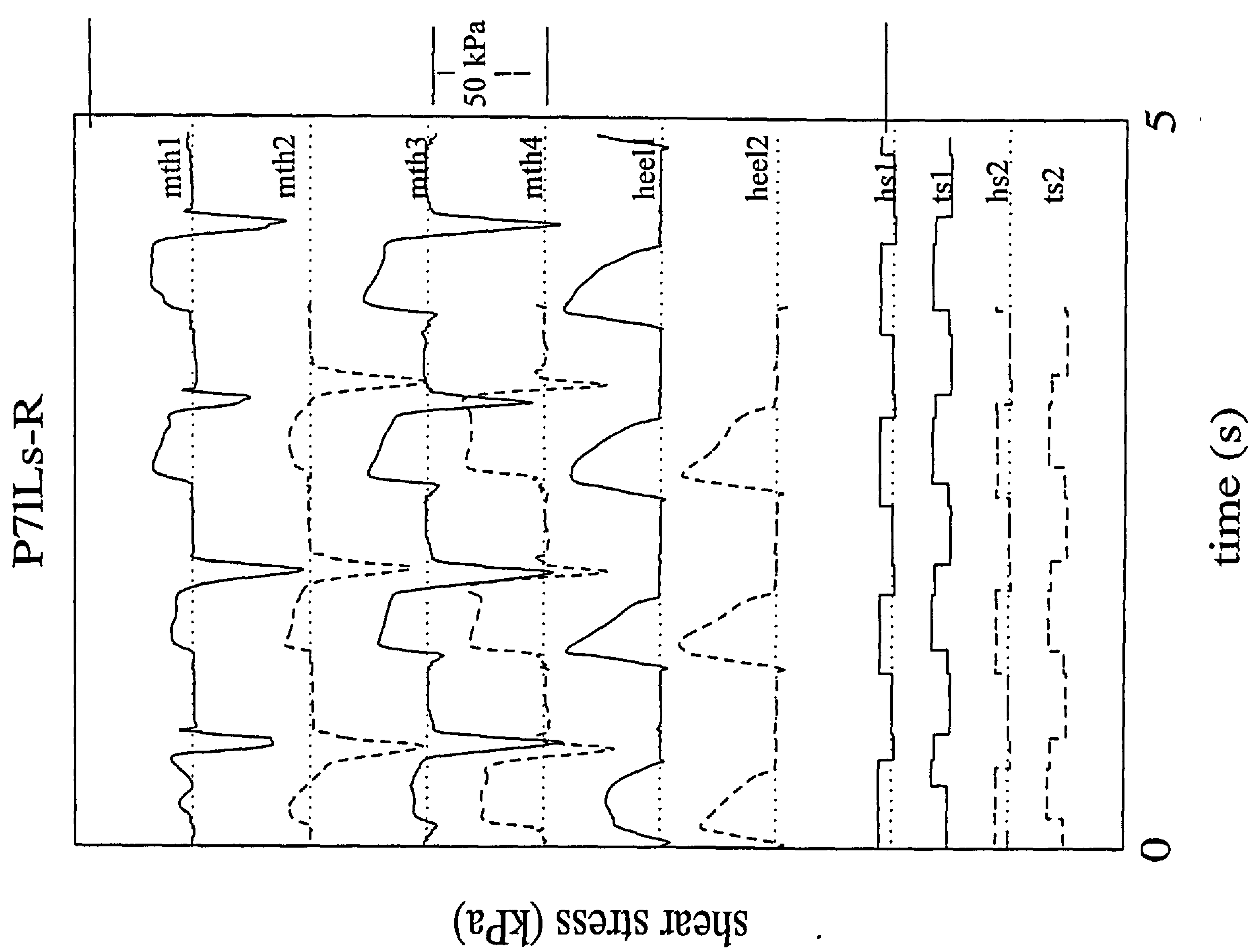

P71Rs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

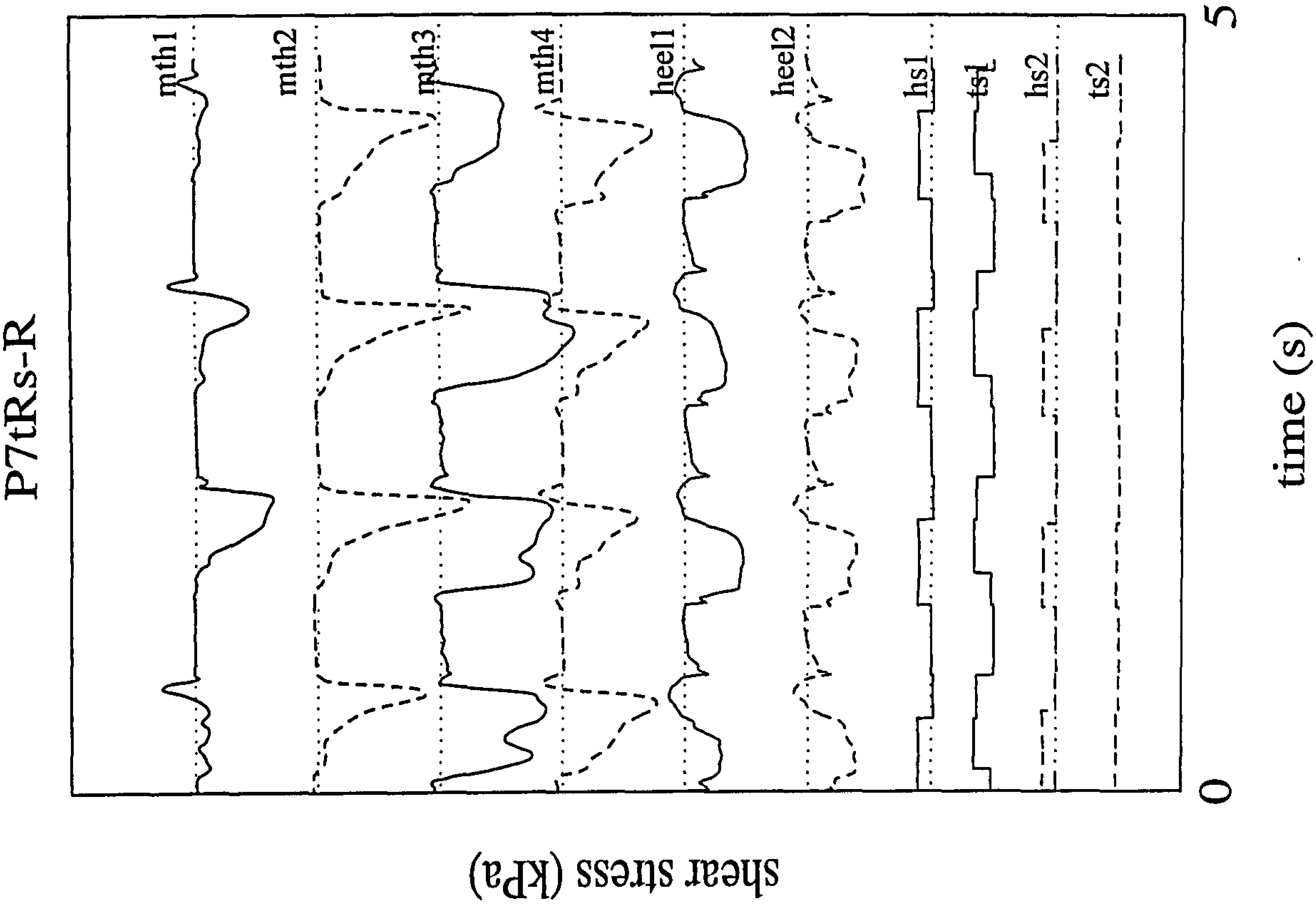

P71Ls
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
shear stress (kPa)
time (s)
0
5

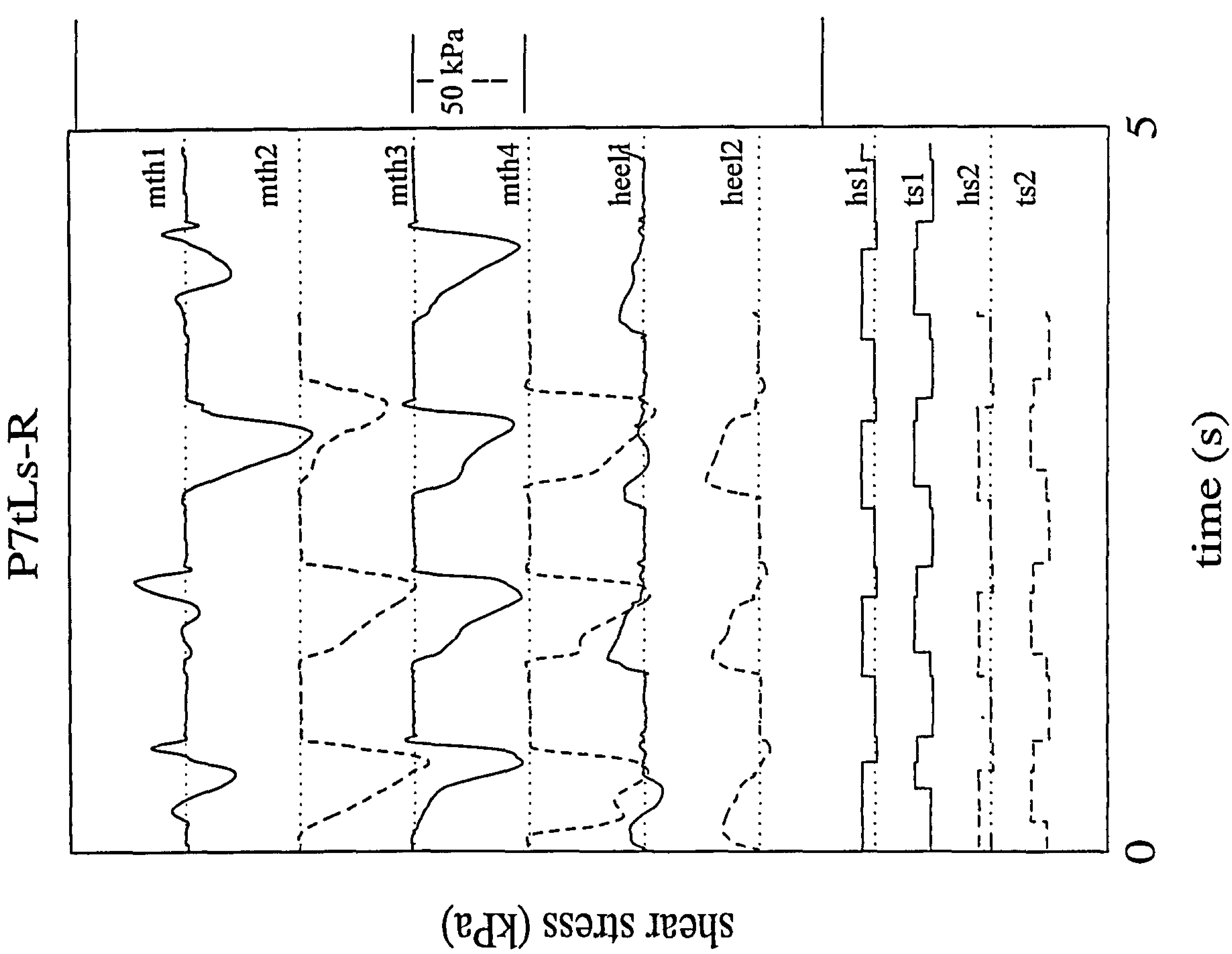

P7tRs
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


P7tLs
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


P71Rs-R
shear stress (kPa)
time (s)
0
5
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2


P71Ls-R
shear stress (kPa)
time (s)
50 kPa
0
5
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2


P7tRs-R
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)


P7tLs-R
50 kPa
mth1
mth2
mth3
mth4
heel1
heel2
hs1
ts1
hs2
ts2
0
5
time (s)
shear stress (kPa)

APPENDIX H

Table H-I. Resultant maximum peak shear stresses (kPa) beneath the feet of the subjects during shod walking (not wearing hosiery).

Subject	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
1	12.6	19.0	26.7	61.8	31.7	58.3	22.6	17,2	50.5	32.8	34.6	28.1
	11.9	21.6	33.4	62.2	44.9	49.5	35.4	13.6	33.5	35.3	44.1	24.7
	12.4	19.3	33.8	56.4	26.8	45.6	20.7	16.4	45.3	43.4	41.6	27.7
2	35.4	32.9	32.3	29.5	32.5	54.2	66.6	21.0	31.2	40.7	29.1	29.9
	34.9	33.4	38.4	32.3	35.0	56.8	73.8	21.8	30.0	33.6	28.6	24.7
	32.1	33.9	40.5	31.8	29.8	58.3	72.0	25.9	30.5	34.5	30.2	26.0
3	11.2	15.3	39.2	27.3	46.4	42.3	11.3	6.3	32.7	38.9	16.2	26.1
	16.2	13.3	34.6	28.5	50.6	41.0	10.3	10.6	40.8	20.4	16.5	20.5
	10.7	11.2	28.2	27.2	44.6	44.6	6.5	8.4	37.8	25.3	15.7	19.3
4	40.9	47.4	59.8	93.7	22.5	46.0	40.9	34.7	116.5	126.5	47.3	39.6
	42.1	46.8	41.0	119.6	18.3	48.2	42.5	33.4	119.2	105.6	54.4	54.0
	41.6	43.9	55.0	107.8	21.6	59.0	42.2	29.0	107.9	104.1	51.8	51.1
5	28.0	19.6	44.8	52.0	26.8	13.2	21.2	11.5	106.7	71.3	44.4	44.1
	32.9	19.7	42.3	56.0	25.6	11.3	26.1	11.2	106.5	64.4	44.2	43.6
	29.0	20.5	45.3	47.0	29.2	19.3	23.5	9.7	111.3	73.8	40.2	37.3
6	114.0	90.0	27.2	33.8	19.9	43.6	35.7	24.4	66.1	42.8	41.8	75.9
	110.6	92.3	25.3	38.3	18.1	39.4	28.2	9.1	48.3	43.5	50.4	89.3
	130.2	109.9	20.5	40.9	19.8	32.6	34.6	9.0	61.3	46.3	38.5	78.5

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.

Table H-I (continued). Resultant maximum peak shear stresses (kPa) beneath the feet of the subjects during shod walking (not wearing hosiery).

Subject	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
7	41.9	37.5	38.2	44.7	17.0	40.2	41.6	59.3	104.3	136.8	78.8	72.9
	44.5	48.7	30.5	49.8	26.8	38.2	59.3	68.5	101.6	125.5	97.3	76.9
	50.7	55.6	41.0	42.4	37.3	53.2	53.1	55.7	107.1	135.8	89.6	84.1
8	37.0	109.5	26.4	70.9	27.4	41.8	31.5	49.8	90.4	87.5	81.1	59.6
	45.3	69.3	26.9	58.8	29.9	46.3	30.3	46.2	83.5	77.0	76.8	79.8
	52.8	77.6	22.6	56.0	33.0	61.9	24.3	52.7	86.2	70.7	78.8	62.6
9	43.1	36.4	54.8	22.3	126.7	67.5	26.0	61.4	181.5	93.9	52.1	39.2
	49.0	40.1	61.6	37.2	122.7	61.0	30.6	64.9	200.5	98.9	40.5	42.7
	66.3	40.8	62.8	32.2	97.4	80.0	30.8	63.8	205.2	108.6	43.8	38.3
9-R	25.6	24.0	17.1	26.6	47.3	40.2	16.8	38.0	95.2	56.8	27.7	40.5
	31.4	27.9	19.0	23.2	51.8	47.2	23.6	42.6	104.4	43.5	24.6	37.6
	22.1	26.0	20.4	17.5	55.4	39.6	21.3	36.8	108.6	54.4	24.8	31.6

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
R - results of the repeat shear test

Table H-II. Resultant maximum peak shear stresses (kPa) beneath the feet of the subjects during shod walking (wearing nylon hold-ups).

Subject	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
1	32.0	23.7	12.4	21.4	10.9	25.2	7.4	10.6	36.7	43.3	24.3	19.9
	35.5	21.6	12.9	26.8	20.8	35.8	7.4	12.9	42.4	38.8	18.9	22.2
	37.2	21.0	13.9	24.2	15.5	40.1	16.7	10.7	26.7	50.7	23.4	27.3
2	31.0	38.3	29.7	35.5	44.9	41.9	31.0	13.0	38.3	37.9	26.3	21.7
	30.9	32.6	34.3	29.8	46.9	45.8	35.7	14.1	34.8	41.0	23.8	23.4
	30.8	30.2	34.3	31.3	49.0	56.6	46.3	14.6	36.6	41.4	21.7	22.0
3	16.2	19.8	12.2	20.7	16.8	14.3	26.4	13.5	27.0	41.9	9.7	16.0
	19.1	21.2	20.4	17.8	17.5	11.6	17.3	16.2	25.3	31.2	9.2	11.3
	20.5	18.8	13.9	21.4	21.2	11.2	13.0	17.1	22.1	40.5	8.5	14.8
4	33.8	42.8	29.5	100.9	29.7	62.3	50.5	35.4	95.2	62.7	51.2	53.0
	31.3	46.0	34.9	103.4	26.3	80.4	52.4	30.6	106.1	77.8	54.8	47.9
	36.2	42.9	34.5	102.3	30.8	82.8	47.4	33.1	101.4	77.7	55.8	48.2
5	28.1	38.4	18.7	57.1	29.6	13.2	27.6	9.8	90.0	63.1	32.1	30.8
	33.9	25.3	31.8	67.7	49.8	9.0	28.2	13.4	91.3	64.7	35.1	32.6
	35.8	24.9	33.8	62.2	19.3	11.3	33.1	19.0	92.3	69.0	45.5	39.5
6	80.3	71.7	20.8	48.5	32.3	33.6	35.0	14.8	51.2	32.8	68.7	60.0
	77.2	51.4	23.8	45.4	21.7	34.6	30.3	8.0	43.0	33.5	72.4	73.2
	82.0	53.5	19.1	41.9	30.9	62.6	51.9	10.1	48.6	27.9	79.2	68.9

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.

Table H-II (continued). Resultant maximum peak shear stresses (kPa) beneath the feet of the subjects during shod walking (wearing nylon hold-ups).

Subject	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
7	47.7	37.7	35.7	47.4	19.8	97.5	94.4	34.6	65.3	83.3	93.0	80.1
	55.1	46.9	23.8	46.4	41.5	124.4	87.7	32.5	59.0	76.6	125.7	99.3
	46.9	42.5	21.2	38.8	28.4	90.9	94.8	26.3	58.0	78.9	126.5	91.1
8	51.4	84.1	15.4	66.6	24.3	80.4	16.9	55.9	120.1	59.3	94.4	83.4
	52.5	74.0	25.6	64.0	31.0	56.0	26.4	50.4	110.4	43.4	101.9	75.1
	53.9	69.9	21.9	70.2	36.8	59.1	27.9	51.7	104.4	69.4	97.6	96.7
9	36.0	45.2	42.1	30.9	73.3	24.9	39.0	37.0	129.1	87.2	33.9	38.4
	45.7	44.0	39.2	48.3	88.4	18.0	39.7	32.5	119.7	75.7	32.2	35.2
	47.8	45.3	41.8	36.6	82.0	27.6	54.9	29.3	125.8	75.5	22.3	44.1
9-R	39.0	34.7	13.3	21.2	33.2	26.3	16.2	28.9	76.9	40.2	25.9	28.7
	38.2	36.3	15.1	30.3	39.6	31.4	16.2	24.7	77.7	37.9	32.8	22.8
	39.5	38.4	15.8	20.8	43.7	32.6	22.9	19.1	66.7	46.2	28.2	21.2

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
R - results of the repeat shear test

Table H-III. Resultant maximum peak shear stresses (kPa) beneath the feet of the patients during shod walking (not wearing hosiery).

Patient	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
1	82.4	46.3	49.6	56.0	56.4	45.0	119.7	97.1	58.4	50.3	62.1	67.4
	84.6	56.8	46.2	66.0	52.7	61.3	126.5	100.0	45.2	43.7	45.0	78.6
	86.2	79.8	63.3	72.7	53.2	84.1	119.1	99.6	53.8	45.7	50.1	65.5
2	9.3	12.5	40.7	43.9	47.7	19.0	49.8	71.0	55.1	40.3	36.2	52.5
	9.2	11.3	42.8	48.2	58.2	12.0	51.1	71.4	61.5	46.4	26.8	44.5
	8.5	14.1	34.0	46.3	50.3	16.3	51.6	75.0	69.8	48.7	61.1	40.8
3†	62.6	77.6	142.4	101.9	78.3	123.4	-	-	-	-	-	-
	54.9	83.3	108.6	117.2	89.9	117.1	-	-	-	-	-	-
	62.0	70.2	107.9	110.0	39.9	174.4	-	-	-	-	-	-
4	8.1	11.7	77.8	35.7	25.5	46.0	53.3	17.2	52.2	20.9	21.6	27.0
	7.8	10.0	70.8	39.7	19.4	32.6	100.4	24.6	28.2	21.8	22.9	29.1
	10.6	17.6	83.8	36.0	25.9	16.9	62.9	29.4	33.7	23.0	22.5	32.7
5	62.2	51.1	15.6	30.2	52.7	54.1	84.9	22.1	42.9	44.5	35.7	21.4
	54.2	52.1	13.7	30.0	52.4	48.7	93.5	33.0	39.8	45.2	30.6	20.6
	63.0	57.3	24.4	34.3	54.9	55.3	78.6	37.2	42.3	42.1	38.4	25.6

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
† shear stresses were not recorded beneath the right foot

Table H-III (continued). Resultant maximum peak shear stresses (kPa) beneath the feet of the patients during shod walking (not wearing hosiery).

Subject	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
6	29.5	47.2	24.4	19.6	35.3	26.2	58.5	43.0	30.3	28.3	46.8	29.5
	26.5	42.1	19.4	14.9	35.3	18.0	57.2	44.6	26.6	30.5	36.2	34.7
	32.9	26.9	19.2	15.0	46.0	23.8	35.4	37.3	25.7	24.7	25.9	31.7
7	46.7	71.2	63.9	35.4	33.6	41.5	51.0	111.8	83.2	52.6	49.6	63.0
	71.7	90.7	100.9	32.8	39.4	55.8	60.2	118.2	71.4	46.3	75.8	62.3
	67.8	83.6	111.0	33.0	35.8	45.9	53.8	112.2	88.5	53.8	49.3	76.9
7-R	32.1	31.5	74.7	43.8	33.1	23.8	47.3	41.6	55.9	51.5	63.0	47.5
	21.8	44.3	74.9	41.4	30.6	41.2	62.3	66.8	40.6	44.1	74.0	49.9
	24.4	36.8	79.9	42.3	29.2	35.0	60.4	59.8	65.4	42.6	60.4	55.9

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
R - results of the repeat shear test

Table H-IV. Resultant maximum peak shear stresses (kPa) beneath the feet of the patients during shod walking (wearing nylon hold-ups).

Patient	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
1	69.1	74.3	32.1	60.7	36.6	35.1	93.0	61.7	61.4	45.1	54.6	56.5
	69.7	76.5	29.1	41.9	65.0	28.9	86.8	49.3	62.8	20.3	52.4	70.4
	69.2	75.1	26.2	45.5	63.2	32.9	86.0	66.4	65.7	46.3	45.1	66.7
2	9.4	13.3	27.4	31.9	22.3	22.9	46.5	63.0	43.2	47.5	31.5	27.2
	8.6	5.4	35.4	43.3	26.0	27.4	48.5	65.6	42.5	42.1	39.0	40.1
	10.8	10.0	30.0	42.3	21.7	25.4	26.4	57.6	66.4	43.3	41.1	25.7
3†	49.1	55.1	96.8	119.2	114.5	150.7	-	-	-	-	-	-
	47.8	55.5	90.0	85.0	113.3	143.4	-	-	-	-	-	-
	43.3	51.2	82.2	101.7	103.8	113.8	-	-	-	-	-	-
4	21.2	10.5	82.8	29.6	13.8	27.3	48.6	36.5	53.5	28.4	19.6	41.9
	14.1	7.4	80.7	44.3	17.0	18.2	51.5	34.7	46.3	40.0	18.1	32.5
	23.9	8.1	78.5	40.8	16.6	18.4	74.8	36.5	36.9	32.3	30.3	29.2
5	65.0	50.8	14.4	41.4	66.1	34.8	44.8	15.8	42.3	40.9	38.7	15.2
	54.2	49.0	13.4	22.5	64.8	29.2	44.2	27.1	45.3	41.1	31.4	20.6
	74.5	61.7	16.8	42.8	74.8	37.4	46.8	24.4	42.7	38.4	33.1	18.9

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
† shear stresses were not recorded beneath the right foot

Table H-IV (continued). Resultant maximum peak shear stresses (kPa) beneath the feet of the patients during shod walking (wearing nylon hold-ups).

Subject	Left foot						Right foot					
	heel(1,3)	heel(2,4)	mth4	mth3	mth2	mth1	mth1	mth2	mth3	mth4	heel(1,3)	heel(2,4)
6	24.1	30.9	22.9	53.2	38.4	43.0	60.1	41.1	28.6	44.5	32.7	35.9
	38.2	39.8	15.5	43.4	43.2	35.1	56.8	50.7	31.1	36.1	28.6	25.3
	41.1	30.3	14.2	42.2	37.3	30.4	42.3	42.5	27.5	28.8	33.4	32.4
7	54.2	67.5	44.7	41.7	31.0	50.3	58.2	135.7	38.4	57.4	54.0	45.5
	62.8	93.2	70.3	45.0	36.5	71.8	65.3	119.2	55.8	28.0	48.3	60.6
	63.4	89.1	71.7	37.1	40.6	71.8	56.2	109.2	59.6	40.5	42.7	57.0
7-R	23.9	36.0	57.0	67.5	66.5	36.7	36.9	45.5	65.7	48.0	36.8	52.3
	43.1	46.3	61.9	64.5	59.5	50.7	63.8	64.1	67.3	33.8	37.7	54.3
	38.9	46.9	65.4	47.5	56.4	56.3	55.0	80.2	90.8	36.4	40.4	56.4

heel(1,3) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the first and third metatarsal heads.
heel(2,4) - shear stresses recorded beneath the heel at the same time shear stresses were recorded beneath the second and fourth metatarsal heads.
R - results of the repeat shear test

LIST OF PUBLICATIONS

The research described in this Thesis has led to the publication and presentation of the following papers and conference abstracts.

Papers

Lord M, Hosein R, Williams RB

Method for in-shoe shear stress measurement. J Biomed Eng 1992;14(3):181-186.

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